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This research investigated the collective interactions between hamstrings extensibility, anterior knee laxity, and hamstrings and quadriceps muscle activation as predictors of anterior knee shear forces during single-leg landings in females. Forty-five healthy, recreationally active females participated in a single data collection session during which hamstrings muscle extensibility and anterior knee laxity were measured, followed by measurement of surface electromyography of the quadriceps and hamstrings and estimation of anterior knee shear forces during single-leg landings. Five single-leg drop landings were conducted from a 30 cm platform positioned 30% of the height of the participant behind the center of a force plate. Electromyographic data were normalized to maximum volitional isometric contractions at 30° of knee flexion, and forces were normalized to body weight. Multiple linear regression analyses were used to evaluate the ability of hamstrings extensibility, anterior knee laxity, and hamstrings and quadriceps pre- and post-landing activation amplitudes to predict initial, rate, and peak anterior knee shear forces during the landings. The primary findings were that hamstrings pre-landing activation negatively predicted anterior knee shear force at initial ground contact and positively predicted the rate of anterior knee shear force development following landing. Furthermore, peak anterior knee shear force following the landings was positively predicted by hamstrings post-landing activation and negatively predicted by hamstrings pre-landing activation. Anterior knee laxity, hamstrings extensibility, and quadriceps pre- and post-landing activations did not significantly add to the prediction of

anterior knee shear forces. Hence, it was concluded that hamstrings activation was the primary predictor of anterior knee shear forces during single-leg landings in females.

ANATOMICAL AND NEUROMUSCULAR CONTRIBUTIONS TO ANTERIOR KNEE SHEAR FORCE DURING SINGLE-LEG LANDINGS IN FEMALES

By

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Approved by

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To my sister Lynn,

You have always supported me with your unconditional love.

and to my Mom and Dad,

You have encouraged me to chase my dreams and always given me the support necessary to achieve them.

and most of all to my wife Jen,

Your constant love, support, and belief in me are my inspiration in life.

In honor of my grandparents, Helen Mary Hertrich, Elgarie Windley, William Windley, and in loving memory of my grandfather, Frederick Hertrich Jr.

APPROVAL PAGE

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CHAPTER I

INTRODUCTION

Over the past two decades increased female participation in sport has highlighted sex discrepancies in injuries to the anterior cruciate ligament (ACL) of the knee. Female athletes injure their ACLs at rates from two to eight times greater than male athletes during sports that involve cutting, stopping, and landing (Agel, Arendt, & Bershadsky, 2005; Arendt, Agel, & Dick, 1999; Arendt & Dick, 1995; Biondino, 1999; Ferretti & Papandrea, 1992; Malone, Hardaker, Garrett, Feagin, & Bassett, 1993; Oliphant & Drawbert, 1996). ACL injuries to high school and college female athletes represent greater than one-third of the incidence in the United States population and of the estimated $1.4 billion spent each year in this country on surgical and rehabilitative costs alone (Henry & Kaeding, 2001; Miyasaka, Daniel, Stone, & Hirschman, 1991). This estimate does not account for the hardships or costs resulting from the long term debilitating problems leading to osteoarthritis that most females with ACL injury will have to endure later in life (Griffin et al., 2000; Lohmander, Ostenberg, Englund, & Roos, 2004; McAllister et al., 2003; Murrell, Maddali, Horovitz, Oakley, & Warren, 2001; von Porat, Roos, & Roos, 2004). The large sex discrepancies as well as the immediate and long term implications of ACL injury necessitate research that examines the risk factors and mechanisms associated with this injury.

A large percentage of ACL injuries are non-contact in nature (48 to 96%)(Agel et al., 2005; Arendt & Dick, 1995; Boden, Dean, Feagin, & Garrett, 2000; Ferretti & Papandrea, 1992; Myklebust, Maehlum, Engebretsen, Strand, & Solheim, 1997; Olsen, Myklebust, Engebretsen, & Bahr, 2004) and occur during landing (37 to 73%) (Boden et al., 2000; Ferretti & Papandrea, 1992). Of those that occur during landing, the majority of the weight is typically on one leg (Olsen et al., 2004) with the knee positioned in a small flexion angle (McNair, Marshall, & Matheson, 1990; Olsen et al., 2004). ACL injury mechanism theory indicates that, in this position, the collective biological systems are not able to control the knee joint in the sagittal plane resulting in injury to the ACL (Boden et al., 2000; Hewett, Stroupe, Nance, & Noyes, 1996; Huston & Wojtys, 1996; Ireland, 1999). Due to the likelihood of sustaining an ACL injury during landing via this mechanism (Boden et al., 2000; Ferretti & Papandrea, 1992), landing has been suggested as ideal for the analysis of dynamic sagittal plane knee joint stabilization (Decker, Torry, Noonan, Riviere, & Sterett, 2002).

For the purpose of this dissertation, sagittal plane knee joint stabilization refers to the collective and interactive role of the anatomical structures (eg. ligaments, tendons, joint articulations) and neuromuscular system in controlling the motion and forces at the knee joint in the sagittal plane. While direct measurement of the loads imposed on the ACL may not be practical during a landing study, anterior knee shear force (AKSF) may be an indicator of sagittal plane knee joint stabilization. AKSF has been proposed as a risk factor for ACL injury (Chappell, Yu, Kirkendall, & Garrett, 2002), and when applied externally, has been shown to result in increased anterior tibia translation (Beynnon et al., 1997; DeMorat, Weinhold, Blackburn, Chudik, & Garrett, 2004; Fleming, Brattbakk, Peura, Badger, & Beynnon, 2002), ACL strain (Berns, Hull, & Patterson, 1992; Beynnon et al., 1997; Fleming, Renstrom, Beynnon et al., 2001), and ACL tension (Butler, Noyes, & Grood, 1980; Chan & Seedhom, 1999; Gabriel, Wong, Woo, Yagi, & Debski, 2004; Markolf et al., 1995; Markolf, Gorek, Kabo, & Sharpiro, 1990; Sakane et al., 1999) in humans. When estimated with the inverse dynamics solution during motion analysis, anterior knee shear force (AKSFid) represents the summed effect of all structures that produce forces across the knee joint in the sagittal plane (Robertson, Caldwell, Hamill, Kamen, & Whittlesey, 2004). While it is not possible to measure the musculo-tendinous and ligamentous forces directly with this model (Winter, 1990), a close examination of this model indicates increases in posterior ground reaction force and quadriceps muscle activations may increase AKSFid while hamstrings muscle activations may decrease AKSFid. Using this model, females have been reported to sustain greater peak AKSFid than males during landings (Chappell et al., 2002), however investigations are limited and have yet to explain the underlying causes for the greater AKSFid seen in females.

Theoretical concepts and research suggest that an interaction between sex specific anatomical characteristics and the efficiency of neuromuscular activation patterns in the lower extremity may, in part, explain this phenomenon. Hamstrings muscle extensibility is generally greater in females than males (Blackburn, Riemann, Padua, & Guskiewicz, 2004) and, when high, may reduce the ability of the muscle-tendon unit to generate stiffness (Blackburn, Padua, Riemann, & Guskeiwicz, 2004) and force (Chow, Medri, Martin, Leekam, & McKee, 2000). Anterior knee laxity is also greater in females

(Huston & Wojtys, 1996; Rosene & Fogarty, 1999; Rozzi, Lephart, Gear, & Fu, 1999; Trimble, Bishop, Buckley, Fields, & Rozea, 2002), and females with increased anterior knee laxity have demonstrated greater lateral hamstrings activation levels when exposed to knee loading (Shultz, Carcia, Gansneder, & Perrin, 2004). Hence, females with greater hamstrings extensibility may have a reduced ability to resist the development of AKSFid due to a decrease in the hamstrings muscle efficiency, while females with greater anterior knee laxity may increase the activation amplitudes of the hamstrings to reduce AKSFid. It is still unclear how the interactions between hamstrings extensibility, anterior knee laxity, and neuromuscular activation amplitudes of the hamstrings and quadriceps may influence the ability to control AKSFid during a landing in females.

Purpose

The purpose of this study was to investigate the collective interactions between hamstrings extensibility, anterior knee laxity, and hamstrings and quadriceps muscle activation as predictors of anterior knee shear forces during single-leg landings in females.

Hypotheses

Hypothesis 1: Hamstrings extensibility, anterior knee laxity, and hamstrings pre-landing (100ms prior to landing) activation amplitude will negatively predict, and quadriceps pre-landing activation amplitude will positively predict AKSFid at initial ground contact (40N of vertical GRF) of a single-leg landing.

H_0: $\beta_{HS} = 0$, $\beta_{KT} = 0$, $\beta_{Hpre} = 0$, $\beta_{Qpre} = 0$

H_1: $\beta_{HS} < 0$, $\beta_{KT} < 0$, $\beta_{Hpre} < 0$, $\beta_{Qpre} > 0$

Hypothesis 2: Anterior knee laxity and hamstrings pre- and post-landing (100ms following initial ground contact) activation amplitude will negatively predict, and hamstrings extensibility and quadriceps pre- and post-landing activation amplitude will positively predict the rate of AKSFid development following a single-leg landing.

H_0: $\beta_{HS} = 0$, $\beta_{KT} = 0$, $\beta_{Hpre} = 0$, $\beta_{Qpre} = 0$, $\beta_{Hpost} = 0$, $\beta_{Qpost} = 0$

H_1: $\beta_{KT} < 0$, $\beta_{Hpre} < 0$, $\beta_{Hpost} < 0$, $\beta_{HS} > 0$, $\beta_{Qpre} > 0$, $\beta_{Qpost} > 0$

Hypothesis 3: Anterior knee laxity and hamstrings pre- and post-landing activation amplitude will negatively predict, and hamstrings extensibility and quadriceps pre- and post-landing activation amplitude will positively predict the peak AKSFid following a single-leg landing.

H_0: $\beta_{HS} = 0$, $\beta_{KT} = 0$, $\beta_{Hpre} = 0$, $\beta_{Qpre} = 0$, $\beta_{Hpost} = 0$, $\beta_{Qpost} = 0$

H_1: $\beta_{KT} < 0$, $\beta_{Hpre} < 0$, $\beta_{Hpost} < 0$, $\beta_{HS} > 0$, $\beta_{Qpre} > 0$, $\beta_{Qpost} > 0$

Assumptions and Delimitations

1. Only females were included in this study in an effort to eliminate other potential sex-confounding factors on the relationships being examined.
2. Only healthy, recreationally active females between the ages of 18 and 30 were studied.

3. Females were in the first eight days of their menstrual cycle during data collection to avoid hormonal effects on knee laxity.

4. Only females who participate in recreational exercise for at least 90 minutes per week were included.

5. Only a landing task was studied.

6. Anterior knee laxity is a reliable and valid measure of the amount of anterior translation of the tibia relative to the femur.

7. Hamstrings extensibility is a reliable measure of the ability of the hamstrings to elongate.

8. Anterior knee shear force as estimated with the inverse dynamics solution is a valid and reliable measure of the summed effect of all the forces acting on the knee joint in the sagittal plane.

9. Electromyographic amplitude is an adequate representation of the amount of muscle activation.

Limitations

1. The results of this dissertation can not be generalized to populations other than healthy, recreationally active females between the ages of 18 and 30.

2. ACL strain was not directly measured in this study and the calculation of AKSFid is only a model to estimate shear force at the knee.

3. ACL injury risk was not measured in this study.

4. The regression analyses used to represent the models to estimate AKSFid are only evident of the predictor variables included in them and do not indicate how other alignment, biomechanical, or neuromuscular factors may interact with these variables to influence AKSFid.

5. The results of this study cannot be generalized to tasks other than single-leg landings.

Operational Definitions

Healthy – no history of injury to either lower extremity in the past 6 months that has limited normal activities for more than one day; no previous history of ligament rupture in either lower extremity; no previous history of surgery to either lower extremity; and no medical conditions that prohibit current exercise participation.

Recreationally active – current participation in a minimum of 90 minutes of recreational exercise per week and no current participation in professional or intercollegiate athletics.

Preferred leg – the leg on which the participant lands in two of the first three practice landings.

Externally applied anterior knee shear force – an externally applied anterior force of the proximal tibia or shank on the distal femur or thigh or a posterior force of the distal femur or thigh on the proximal tibia or shank.

Anterior knee shear force as estimated with the inverse dynamics solution (AKSFid) – an anterior shear force of the proximal tibia or shank relative to the femur or thigh or a posterior shear force of the distal femur or thigh relative to the proximal tibia or shank as estimated by the inverse dynamics solution divided by the weight (in Newtons) of the participant [expressed in units of body weight (BW)].

Initial contact of landing – the first frame of data when the vertical ground reaction force is greater than or equal to 40N.

Maximum volitional isometric contraction (MVIC) – the average peak muscle sEMG signal amplitude across the middle three of five seconds of a maximum contraction of the quadriceps (vastus lateralis) or hamstrings (semimembranosis and semitendinosis, biceps femoris) as generated by the participant with the knee fixed at 30 degrees of flexion.

Normalized activation amplitude (%MVIC) – the sEMG signal amplitude of each muscle prior to or during the landing task divided by the MVIC for that muscle.

Predictor Variables

Hamstrings extensibility (HS) – the sagittal knee angle (°) with the participant supine, the hip flexed to 120°, and the participant actively extending the knee (greater angle = greater extensibility).

Anterior knee laxity (KT) – passive anterior displacement of the tibia relative to the femur (mm) when 133N of anteriorly directed force is applied to the posterior tibia with the femur stabilized and the participant positioned supine with the knee in 25° to 30° of flexion.

Hamstrings pre-landing activation amplitude (H_{pre}) – the normalized mean of the medial and lateral hamstrings sEMG signal amplitude during the last 100ms prior to initial contact.

Hamstrings post-landing activation amplitude (H_{post}) – the normalized mean of the medial and lateral hamstrings sEMG signal amplitude during 100ms following initial contact.

Quadriceps pre-landing activation amplitude (Q_{pre}) – the normalized mean of the vastus lateralis sEMG signal amplitude during the last 100ms prior to initial contact.

Quadriceps post-landing activation amplitude (Q_{post}) – the normalized mean of the vastus lateralis sEMG signal amplitude during 100ms following initial contact.

Dependent Variables

Initial anterior knee shear force (iAKSF) – the AKSFid value at the time of initial contact of landing (expressed in units of BW).

Peak anterior knee shear force (pAKSF) – the greatest ASKFid value obtained following initial contact of the landing (expressed in units of BW).

Rate of anterior knee shear force (rAKSF) – the pAKSF minus the iAKSF divided by the time to pAKSF [expressed in units of BW per second (BW/s)].

CHAPTER II

LITERATURE REVIEW

ACL Injury

Incidence and Costs

Over the past two decades female participation in sport has increased dramatically. Female participation in all NCAA athletics more than doubled from the 1981-82 (74,239) to the 2002-03 seasons (160,650), with soccer having a more than 10-fold increase in the number of females participating (NCAA, 2002-2003). Accompanying this influx of females into sport was evidence of sex discrepancies in injury rates. In particular, female athletes reportedly sustain injury to the anterior cruciate ligament (ACL) of the knee at rates from two to eight times greater than their male counterparts (Agel et al., 2005; Arendt et al., 1999; Arendt & Dick, 1995; Biondino, 1999; Ferretti & Papandrea, 1992; Malone et al., 1993; Myklebust, Maehlum, Holm, & Bahr, 1998; Oliphant & Drawbert, 1996). Specifically, NCAA injury surveillance system data from 1990 to 2002 indicate that female soccer and basketball players are over three times more likely and female basketball players are over three and a half times more likely to sustain an ACL injury than male soccer and basketball players (Agel et al., 2005). This risk ratio corresponds to one in 40 basketball and one in 48 soccer NCAA female athletes sustaining an ACL injury per year (NCAA, 2003).

The cost of ACL injuries has both immediate and future impact on the individual and society. ACL injuries have been estimated to occur in approximately 80,000 Americans each year (Miyasaka et al., 1991), with a conservative cost estimate of $17,000 per injury in surgical and rehabilitative costs (Henry & Kaeding, 2001) that result in an estimated annual economic impact of approximately $1.4 billion. Greater than one-third of these injuries (approximately 30,000) occur each year in female high school and college athletes, despite this group representing less than one percent of the population of the United States (Henry & Kaeding, 2001). The resulting annual economic impact of these injuries does not take into account the cost of the treatment of an ACL injury prior to surgery, or rehabilitation that results from long-term debilitating problems associated with ACL injury (Griffin et al., 2000). Specifically, long term implications for the majority of those who suffer ACL injury include reduced activity levels (von Porat et al., 2004), knee pain (Lohmander et al., 2004; McAllister et al., 2003), cartilage loss (Murrell et al., 2001) and osteoarthritis later in life (Lohmander et al., 2004; von Porat et al., 2004). Hence, the alarming number of females suffering this devastating injury and the resulting immediate and long term economic impacts illustrate the importance of research endeavors which aim to better understand the mechanisms and risk factors associated with ACL injury.

Mechanism of Injury

A large percentage (48 to 96%) of ACL injuries occur in the absence of contact with another athlete or object (Agel et al., 2005; Arendt & Dick, 1995; Boden et al., 2000; Ferretti & Papandrea, 1992; Myklebust et al., 1997; Olsen et al., 2004). The activities at the time of injury most often involve some type of cutting, sudden deceleration, or landing maneuver (Arendt et al., 1999; Arendt & Dick, 1995; Biondino, 1999; Ferretti & Papandrea, 1992; Harmon & Dick, 1998; Malone et al., 1993; Olsen et al., 2004), with landing from a jump reported as the mechanism for 37% (Boden et al., 2000) to 73% (Ferretti & Papandrea, 1992) of these injuries. Observations indicate that a large percentage of these non-contact injuries to the ACL occur with all of the weight on one leg (79%) (Olsen et al., 2004) and the knee in a small flexion angle (90%) (McNair et al., 1990; Olsen et al., 2004).

These data have lead to two main theories for the mechanism of ACL injury both of which describe an event where the majority of the weight is on one foot and the knee is in a small flexion angle (Boden et al., 2000; Hewett et al., 1996; Huston & Wojtys, 1996; Ireland, 1999). The "position of no return" has been described as an uncontrolled landing in sport in which muscles of the lower extremity are unable to control the position of the joints of the lower extremity with a knee positioned in slight flexion (Ireland, 1999). Others have described a "quadriceps dominant" muscle activation pattern in which females land with the knee near full extension while an eccentric quadriceps contraction produces an anterior force on the tibia relative to the femur that is sufficient to overcome the protective posterior force on the tibia produced by the hamstrings (Boden et al., 2000; Hewett et al., 1996; Huston & Wojtys, 1996). The end result is anterior tibial translation and tightening of the ACL causing a tear (Boden et al., 2000; Hewett et al., 1996; Huston & Wojtys, 1996). This theory is supported by a two-dimensional synthetic knee model that indicates all fibers in the ACL may fail simultaneously at a transition angle (5 to 20°)

of knee flexion when the tibia is subjected to an anterior shear force relative to the femur (Zavatsky & Wright, 2001). While these are only theories, they support the need to evaluate the collective biological systems as they contribute to stabilizing the knee in the sagittal plane during dynamic activities such as landing.

Sagittal Plane Knee Joint Stabilization

Given the prevalence of ACL injury during landing (Boden et al., 2000; Ferretti & Papandrea, 1992) and the potential to injure the ACL with sagittal plane forces (Zavatsky & Wright, 2001), landing has been suggested as ideal for the analysis of high velocity dynamic sagittal plane knee joint stabilization (Decker et al., 2002). For the purpose of this dissertation, sagittal plane knee joint stabilization refers to the collective and interactive role of the anatomical structures (eg. ligaments, tendons, joint articulations) and neuromuscular system in controlling the motion and forces at the knee joint in the sagittal plane. In order to select and interpret measures of sagittal plane knee stability during landing, a thorough understanding of the factors that contribute to sagittal plane knee stabilization is necessary. These factors are anatomical joint congruence and the passive and active structures that reside within and about the knee joint.

Joint Congruence

The knee joint is markedly incongruent and is formed by articulations between the convex surfaces of the condyles of the distal femur and the concave surfaces of the condyles of the proximal tibia (Chan & Seedhom, 1999). The superior surfaces of the tibial condyles are covered by medial and lateral meniscal discs that distribute contact forces and reduce friction in the knee joint (Chan & Seedhom, 1999). Under axial loading, these menisci have been observed to limit anterior translation of the tibia relative to the femur and tension on the ACL when the knee is exposed to an anteriorly directed force (Chan & Seedhom, 1999; Shoemaker & Markolf, 1986). The contribution of these structures to resist sagittal plane knee joint translation has been reported to range from 12.3% at zero degrees of knee flexion to 22.3% at 30° of knee flexion (Sakane et al., 1999). These data suggest that, while the bony knee geometry may play a role in controlling the knee in the sagittal plane, that role appears to be minimal.

<u>Passive Structures</u>

Due to the relative tibio-femoral joint incongruence, the passive structures both within and surrounding the knee joint must also contribute to sagittal plane knee joint stabilization (Sakane et al., 1999). The passive structures include a joint capsule and four ligaments that guide the joint motion and resist loads applied to the joint that would otherwise cause episodes of instability. Of these passive structures, the primary contributor to stabilization of the knee under an anteriorly directed load is the ACL (Butler et al., 1980; Chan & Seedhom, 1999; Fleming, Renstrom, Beynnon et al., 2001; Markolf et al., 1990; Moglo & Shirazi-Adl, 2003; Sakane et al., 1999). The ACL resists up to 86% of anterior translation of the tibia relative to the femur with no joint compression (Butler et al., 1980) and up to 81% with a knee joint compression equal to twice body weight (Chan & Seedhom, 1999).

Active Structures

Active structures that cross the knee play a major role in controlling joint motion during dynamic function. The major active force generating structures that cross the knee are the quadriceps, hamstrings, and gastrocnemius muscles. Prior to a landing, these muscles are pre-activated to prepare for joint loading (Cowling & Steele, 2001; Fagenbaum & Darling, 2003), and as weight is accepted a vertical ground reaction force causes the hip, knee, and ankle to flex (Decker et al., 2002; Decker, Torry, Wyland, Sterett, & Steadman, 2003; Tillman, Criss, Brunt, & Hass, 2004). While the quadriceps are active to slow knee flexion eccentrically (Tillman et al., 2004), the force generated through the patellar tendon also causes an increase in anterior translation of the tibia relative to the femur (Hirokawa, Solomonow, Luo, Lu, & D'Ambrosia, 1991; Li et al., 1999). Concurrently, the hamstrings are activated primarily to slow hip flexion (Tillman et al., 2004), and the force generated through the distal hamstrings tendons reduces anterior tibia translation (Hirokawa et al., 1991; Li et al., 1999). The gastrocnemius also crosses the knee, and early data suggests it may increase ACL strain at low knee flexion angles (Fleming, Renstrom, Ohlen et al., 2001). However, the gastrocnemius is primarily activated to slow dorsiflexion (Tillman et al., 2004), and most likely has a minimal effect on sagittal plane knee joint stabilization. These data suggest that, as the joints of the lower extremity are loaded during a landing, the muscles that cross the knee joint may influence sagittal plane knee joint stabilization through their modulation of the sagittal movements at the knee. While isolated quadriceps muscle contractions increase anterior tibial translation, isolated hamstrings muscle contractions decrease anterior tibia translation, and further investigations are necessary to understand fully the role of the gastrocnemius on sagittal plane knee joint stabilization.

Interaction Between Passive and Active Structures

Based on the active and passive contributions to sagittal knee joint stabilization, the ACL and hamstrings muscle appear to be the primary restraints to anterior tibia translation. Mechanoreceptors have been identified within the human ACL (Adachi et al., 2002), however there is considerable debate regarding a proposed direct synergistic relationship between the ACL and the hamstrings muscle as a mechanism for controlling sagittal plane knee joint stabilization (Baratta et al., 1988; Fujita, Nishikawa, Kambic, Andrish, & Grabiner, 2000; Grabiner, Campbell, Hawthorne, & Hawkins, 1989; Grabiner, Koh, & Miller, 1992; Raunest, Sager, & Burgener, 1996; Solomonow et al., 1987; Tsuda, Okamura, Otsuka, Komatsu, & Tokuya, 2001). Direct loads on the feline, sheep, and goat ACLs have been observed to result in a hamstrings muscle reflex, termed the ACL-hamstrings reflex (Fujita et al., 2000; Raunest et al., 1996; Solomonow et al., 1987). In humans hamstrings muscle EMG activity has been observed to increase during isokinetic quadriceps, leading the investigators to conclude that the activations in the antagonist muscles were to enhance knee stabilization in response to the loads placed on the joint by the agonist (Baratta et al., 1988). Alternately, under isometric conditions hamstrings muscle activation amplitudes have not been reported to increase when quadriceps activation amplitudes increased at 10% intervals (Grabiner et al., 1989) or when knee flexion angles were varied (Grabiner et al., 1992), leading the investigators to concluded that there is no direct ACL-hamstring muscle synergy in humans under isometric loads. While an ACL-hamstrings muscle synergy may not exist under static indirect loading, direct or dynamic loading of the ACL in animals and humans does appear to initiate an ACL-hamstrings reflex arc. Hence, integrity or structural soundness of the ACL not only contributes to passive stability, but may also play an important role in the ability of the hamstrings muscle to contribute actively to sagittal plane knee joint stabilization.

Summary

The knee joint is a markedly incongruent articulation between the femoral and tibial condyles. Due to this incongruence, the passive and active structures that reside within and about the joint play critical roles in contributing to sagittal plane knee stability. The ACL is the primary passive restraint to anterior movement of the tibia relative to the femur in the sagittal plane, and that the quadriceps muscle increases anterior translation of the tibia, while the hamstrings muscle reduces anterior translation of the tibia. Hence, the quadriceps muscle appears to be an antagonist to the ACL while the hamstrings appear to be protective of the ACL, making the hamstrings critical for their contribution to controlling anterior tibia translation in the sagittal plane. A protective interaction between mechanoreceptors in the ACL and the hamstrings muscle appears to augment these stabilization strategies during dynamic loading. This protective interaction suggests that integrity of the ACL is critical to both passive and active contributions to sagittal plane knee stabilization. If all of these systems work in concert, the loads imposed on the knee joint during landing will be shared, and theoretically no injury will be sustained. However, if one or more of these contributions to sagittal plane knee stabilization fail, then excessive strain and injury to the ACL may occur.

The Relationship Between Anterior Knee Shear Force and ACL Strain

Anterior directed knee joint forces have been utilized as estimates of sagittal plane knee joint stabilization during landing (Chappell et al., 2002). However, before anterior directed knee joint forces can be discussed as risk factors for ACL injury it is important to understand first how these forces are defined and the subsequent relationship to loads imposed upon the ACL. Anterior knee shear force (AKSF) is an anterior directed force or load of the proximal tibia on the distal femur (Stuart, Meglan, Lutz, Growney, & An, 1996) or conversely stated, a posterior directed force of the distal femur on the proximal tibia (Moglo & Shirazi-Adl, 2003). When the knee is flexed between 10° and 60°, externally applied AKSFs from 130N to 150N have been observed to increase anterior translation of the tibia relative to the femur by approximately 4mm (Beynnon et al., 1997; Fleming et al., 2002). As anterior tibial translation occurs, the ACL is responsible for up to 81% and 86% of the passive resistance to AKSF with (Butler et al., 1980) and without (Chan & Seedhom, 1999) axial compressive joint loads, respectively. Since it is apparent that an AKSF creates anterior tibia translation that is primarily restrained by the ACL, the end result must be increased strain on the ligament.

When referring to ligament strain, strain is defined as the change in length of a ligament divided by its original length (Woo, Debski, Withrow, & Janaushek, 1999). With the knee in 20° to 30° of flexion, in vivo AKSFs from 130N to 150N have been reported to increase ACL strain in the anteromedial bundle during both non-weight-bearing and simulated weight-bearing by approximately 4% (Fleming, Renstrom, Beynnon et al., 2001). During this investigation applied AKSFs of less than 40N in the weight bearing condition resulted in greater ACL strain than in the non-weight-bearing condition ($P < .01$). Hence, these data suggest that ACL strain may actually be greater in weight bearing as compared to non-weight bearing with similarly applied AKSFs.

The current literature appears to indicate a direct relationship between applied anterior tibia loads (ie. AKSF), increased anterior tibia translation, increased ACL tension, and increased ACL strain. This relationship is such that activities or events that result in significant AKSF may pose a challenge to stabilizing the knee joint in the sagittal plane. It is important to note that these relationships are primarily based on measurement techniques that used externally applied forces (usually in a passive situation not mimicking sporting activity). However, they support the contention that those who experience high AKSFs during weight bearing activity will likely experience periods of high ACL strain, thereby increasing the risk of the ligament tearing (Woo et al., 1999). Hence, estimating the anterior load of the tibia on the femur may help describe the effectiveness of sagittal plane knee joint stabilization in controlling strain on the ACL during functional activities.

Estimation of Anterior Knee Shear Forces During Landing

Before AKSF can be interpreted and given a “real world” context, it is imperative that researchers and clinicians understand how this variable is estimated or modeled during functional activity. During human movement, some variation of the inverse dynamics solution is most commonly used to estimate the sagittal component of the knee joint resultant force, i.e. the shear force which acts perpendicular to the sagittal position of the tibia (AKSFid) (Chappell et al., 2002; Cowling & Steele, 2001; Stuart et al., 1996). Inverse dynamics is a solution that uses participant anthropometrics, segmental kinematics, and ground reaction force data to estimate joint kinetics (Winter, 1990). These estimates are based on the motion of a link-segment model that assumes the body is comprised of links (i.e. joints) that connect several rigid segments (i.e. bony levers). One limitation of using inverse dynamics to estimate joint forces is that it is not cable of calculating the individual muscle or ligament forces occurring at a joint without the use of some optimization technique (Winter, 1990). Therefore, the estimated AKSFid value represents the summed effect of all the structures that produce shear forces across the knee joint perpendicular to the position of the tibia in the sagittal plane (Robertson et al., 2004).

The calculation of ASKFid is performed by solving three equations that together estimate the combined effect of the ground reaction forces and the weight and motion of the segments of the lower extremity on the knee joint. First, the anterior-posterior force at the ankle joint (F_{ax}) is calculated from the anterior-posterior ground reaction force (F_x), the mass of the foot segment (m_f), and the anterior-posterior linear acceleration of the foot segment (a_{fx}) (Robertson et al., 2004).

$$F_{ax} + F_x = m_f a_{fx}$$

Next, the anterior-posterior force at the knee joint (F_{kx}) is calculated from the anterior-posterior ankle force (F_{ax}), the mass of the shank segment (m_s), and the anterior-posterior linear acceleration of the shank segment (a_{sx}) (Robertson et al., 2004).

$$F_{ax} + F_{kx} = m_s a_{sx}$$

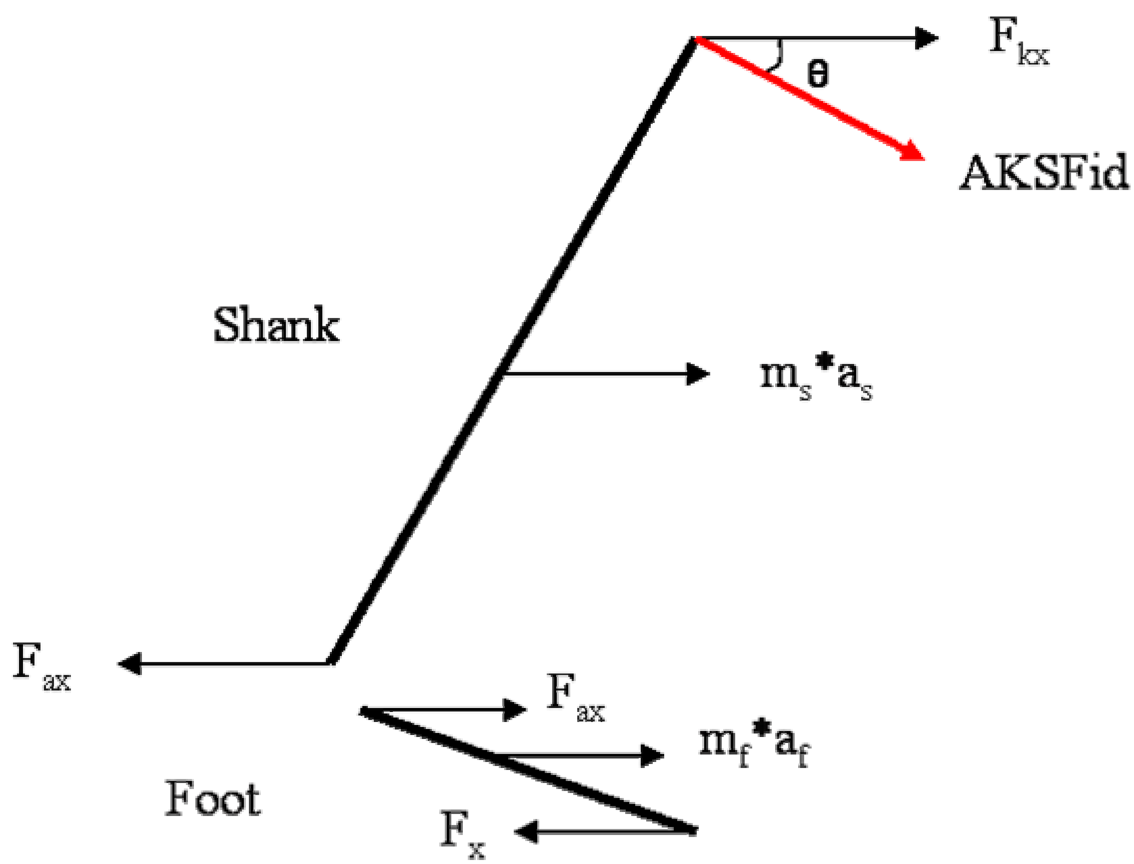
That force is then translated to the shank or tibia reference frame by multiplying the cosine of the angle of the shank with the horizontal (Cosθ) by the anterior-posterior force at the knee joint (F_{kx}).

$$AKSFid = Cos\theta * F_{kx}$$

From these calculations, it is evident that the anterior-posterior ground reaction force, the segmental masses, and the segmental accelerations directly affect the estimation of AKSFid. AKSFid is usually reported as a force of the shank directed perpendicular and anterior to the sagittal position of itself or conversely as a force of the femur directed perpendicular and posterior relative to the sagittal position of the shank (Chappell et al., 2002; Stuart et al., 1996). The free-body diagram used for the inverse dynamics estimation of AKSF is presented in Figure 1.

In order for clinicians to interpret AKSFid data appropriately, it is important to understand the reliability of this measure. Only one peer-reviewed study was found that directly examined the reliability of AKSFid during motion analysis (Chappell et al., 2002). The authors reported coefficients of multiple correlation as estimates of the intra-subject trial repeatability of peak AKSFid and knee extension moments during forward, backward, and vertical stop jump tasks. They reported correlation coefficients >.90 for both variables during each of the three tasks over five consecutive landing trials. While

Figure 1. Free-body Diagram for the Inverse Dynamics Estimation of AKSFid



AKSFid	Anterior knee shear force as estimated by inverse dynamics
F_x	Anterior-posterior ground reaction force
F_{ax}	Anterior-posterior ankle force
F_{kx}	Anterior-posterior knee force
m_f	Mass of the foot segment
m_s	Mass of the shank segment
a_f	Anterior-posterior linear acceleration of the foot segment
a_s	Anterior-posterior linear acceleration of the shank segment

the between day reliability of ASKFid has not been reported, day-to-day reliability estimates for peak anterior-posterior GRF have been reported to range from $ICC_{2,k}$ r = .90 (Diss, 2001) to r = .93 (Ferber, McClay-Davis, Williams, & Laughton, 2002). While the authors are not aware of day-to-day reliability estimates of AKSFid, it appears that motion analysis systems are capable of reflecting consistent estimates of AKSFid between landing trials within day, and the ground reaction force component of the calculation of ASKFid can be consistently measured between days. Investigations that report AKSFid may therefore benefit from reporting its day to day consistency.

Factors that Contribute to the Estimation of AKSFid

Internal factors acting on the proximal and distal segments of the knee joint during movement may influence the estimation of AKSFid. Since this force represents the sum of all sagittal plane forces acting on the knee joint, these internal factors would include the biomechanical and neuromuscular contributions to movement. In reality, the muscle, ligament, and capsular forces imposed on the knee joint are not being independently measured during motion analysis, thus AKSFid represents only a model estimation of the sagittal forces imposed on the knee joint (Winter, 1990). For example, a calculation of AKSFid during quiet stance would indicate little to no force since there is no acceleration of the segments of the knee joint. If, in the same knee flexion angle, the ACL was taught or one were to contract the muscles that cross the knee joint, the resulting calculation of AKSFid would be the same as in the previous example, because there would still be no acceleration of the segments of the knee joint. Hence, it is important to understand that this calculation yields a force that is an estimation of the ability of the collective biological systems to stabilize the knee joint and not an indication of the specific contributions of the biomechanical or neuromuscular factors that influence the segments of the knee joint. The following section will examine the individual ligamentous, biomechanical, and neuromuscular factors that may contribute to or control AKSFid.

Ligamentous

While capsulo-ligamentous tension may not influence the estimation of AKSFid during quiet stance, during movement it is possible that AKSFid is altered by changing tension on the ACL itself. Due to its anatomical attachments on the posterior-lateral femur and the anterior-medial tibia (Moore & Dalley, 1999), the ACL is the primary restraint to anterior translation of the tibia on the femur (Butler et al., 1980), and passive ACL tension may theoretically decrease the anterior acceleration of the tibia, thereby reducing AKSFid. One must be careful when considering this phenomenon, as the tendency may be to consider higher AKSFid to be protective of the ACL during dynamic activity, since the ligament would not be expected to be under enough tension to decrease the AKSFid. When neuromuscular activation is absent, small anterior joint loads may increase ACL strain (Fleming et al., 2002; Fleming, Renstrom, Beynnon et al., 2001), however it seems unlikely during high impact activities that the ACL alone would be able to resist high forces without the contributions of the surrounding musculature. It is also important to remember that the magnitude of the AKSFid is representative of the net of the forces produced by all structures acting on the knee joint in the sagittal plane

(Robertson et al., 2004). A higher AKSFid during dynamic activity, therefore, may indicate that the neuromuscular structures that cross the knee joint may not have activated early enough or with enough force to reduce the anterior acceleration of the tibia, thereby leaving only the ligamentous and capsular restraints to resist the AKSFid. Likewise, a greater AKSFid may indicate that when the ACL becomes taught from anterior movement of the tibia it may be accelerating at such a rate that the force imposed on the ligament may result in injury. Hence, the ACL is only one restraint to the estimated AKSFid and cannot be accurately assessed in isolation using this model.

Biomechanical

According to the inverse dynamics calculation of AKSFid, the biomechanical forces and positions of the lower extremity during landing may influence AKSFid through alterations in the anterior-posterior GRF, sagittal knee position, and the relative efficiency of the lower extremity muscles that cross the knee joint. Theoretically, as the posterior GRF is increased, so will be the ankle joint reaction force in the sagittal plane and subsequently the AKSFid. As the knee is flexed to dissipate the vertical ground reaction force caused by landing (Decker et al., 2002; Decker et al., 2003; McNair & Prapavessis, 1999; Tillman et al., 2004), the relationship between the tibia and the ground changes. Therefore, if the tibia is near perpendicular to the ground, as is the case in a small knee flexion angle, the weight vector will have little to no effect on the acceleration of the tibia as it is oriented nearly perpendicular to the ground. Conversely, with increased knee flexion the tibia is no longer perpendicular to the ground, and the weight vector must be resolved into its anterior and superior components. In this case, the anterior component has a greater contribution to the acceleration of the tibia. Assuming all other variables are constant, the AKSFid would increase slightly as the knee is flexed. Since we know that all other variables do not remain constant in a landing situation, it is likely that the relationship between knee flexion angle and AKSFid is small and is only one contributor to AKSFid during landings.

Neuromuscular

While the direct relationship between knee flexion angle and AKSFid is small, muscle activations during dynamic activity may influence the position, and hence the acceleration of the tibia relative to the femur to modulate sagittal knee stabilization in various knee flexion angles throughout the activity. As an individual accepts weight on the lower extremity during a dynamic task, an AKSFid is imposed on the tibio-femoral joint (Beynnon, Fleming, Labovitch, & Parsons, 2002). The quadriceps muscles are activated prior to and following this weight acceptance to control the resulting knee flexion (Cowling & Steele, 2001). When considered in isolation, a strong quadriceps muscle contraction can increase AKSFid resulting in anterior translation of the tibia at knee flexion angles from 0° to 80° (Hirokawa et al., 1991; Li et al., 1999). If this action is unopposed, this may lead to increased ACL strain, particularly at knee flexion angles from 0° to 45° (Renstrom, Arms, Stanwyck, Johnson, & Pope, 1986). In these lesser knee flexion angles, the quadriceps muscle is in a biomechanically favorable position to produce greater increases in AKSFid, even at low levels of muscle activation, due to its large angle of insertion and longer moment arm distance (Herzog & Read, 1993).

The gastrocnemius and hamstrings are also active prior to and following weight acceptance and may act on the knee to influence AKSFid (Cowling & Steele, 2001). During the deceleration phase of landing, the gastrocnemius slows ankle dorsiflexion, and when considered in isolation, may increase ACL strain from 5° to 15° of knee flexion (Fleming, Renstrom, Ohlen et al., 2001). In contrast, the hamstrings are activated to control hip flexion, and are capable of decreasing anterior tibia translation (Hirokawa et al., 1991; Li et al., 1999) and strain (Renstrom et al., 1986) imposed on the ACL by the quadriceps contraction. However, the effectiveness of the hamstrings in controlling anterior tibia motion is limited to knee flexion angles between 30^{o} and 80°, secondary to their small insertion angle and moment arm from which to generate torque at knee flexion angles less than 30^{o} (Herzog & Read, 1993). These findings are important because they suggest that prior to and immediately following weight acceptance the knee is likely subjected to greater AKSFid in small knee flexion angles secondary to the relative efficiency of the quadriceps and gastrocnemius versus the relative inefficiency of the hamstrings in controlling anterior loads on the knee in the sagittal plane.

Summary

While it is important to note that the relationship between the ligamentous, biomechanical, and neuromuscular factors of landing have not been directly linked to AKSFid during landings, the collective findings suggest that these factors may influence the estimation of AKSFid. Due to the inability of inverse dynamics to estimate ligamentous forces and the impracticality of placing strain transducers in the knees of persons completing landings, it is not possible to estimate how ligamentous tension may influence AKSFid using this model. However, the nature of the calculation does suggest that increases in posterior ground reaction force will result in an increase in AKSFid, and data suggests that increased quadriceps or gastrocnemius activity will increase AKSFid while increased hamstrings activity will decrease AKSFid. Posterior ground reaction forces are readily measured during a landing, and despite the inability of this model to measure muscle forces directly, electromyographic techniques may aid in concurrent estimations of activation levels of the lower extremity muscles. Due to the nature of the calculation of AKSFid, it is not clear how the sagittal knee angle may affect AKSFid during a landing. As the knee is flexed, the neuromuscular efficiency in resisting AKSFid may improve, but the change in the weight vector may effectively cancel out this improved efficiency in controlling AKSFid. Hence, while knee flexion angle should be acknowledged for its possible contribution to AKSFid, it is unclear how it may modulate the inverse dynamics calculation of AKSFid during a landing. The specifics of this model indicate that, during a landing, the posterior ground reaction forces and the quadriceps and hamstrings activation patterns will have the most influence on AKSFid.

Sex-dependent Factors that May Contribute to AKSFid During Landing

Due to the high sex discrepancy in ACL injury, a variety of sex-dependent risk factors have been proposed to explain the higher rates of ACL injury in females. Intrinsic risk factors have been defined as pre-existing internal characteristics and include hormonal fluctuations, joint laxity, limb alignment, femoral notch dimensions, and ligament size (Griffin et al., 2000; McClay Davis I, 2003; McClay-Davis & Ireland, 2001). Extrinsic risk factors have been suggested to be parts of, or responses to, the surrounding environment and include body movement in sport, muscular strength and coordination, landing techniques, and equipment (Griffin et al., 2000; McClay Davis I, 2003; McClay-Davis & Ireland, 2001). While evaluations of these individual risk factors are important groundwork to understand which factors may be sex dependent or occur in correlation with ACL injury, they only describe individual sex differences and do not illustrate how and if they interact with other risk factors to contribute to ACL injury. The most recent consensus statement regarding ACL injury risk factor research highlighted the importance for multi-factoral studies as a more widely faceted approach to understand why females continue to incur ACL injury at a greater rate than males (McClay Davis I, 2003). Interestingly, interactions have been observed to exist between intrinsic and extrinsic risk factors (Heiderscheit, Hamill, & Caldwell, 2000; McClay I, 1998; Shultz, Carcia et al., 2004; Stergiou & Bates, 1997), but it is still unknown how these factors may interact to control knee joint neuromechanics in the sagittal plane during sporting tasks.

Anterior Knee Shear Force as Estimated by Inverse Dynamics

AKSF is an extrinsic risk factor for ACL injury, and female recreational athletes have been reported to experience greater peak AKSFid than male recreational athletes during the landing phase (initial contact to the first local minimum of vertical GRF ~ 50ms) of forward, vertical, and backward stop-jump landings (Chappell et al., 2002). Recent research has supported these findings during a vertical stop jump (Sell et al., 2004) and a single-leg landing (Sander et al., 2004). Consequently, it appears that females may experience greater AKSFid during landing maneuvers, but how these shear forces may be predicted and why these shear forces are occurring are still in question. While investigations regarding sex differences in anterior-posterior ground reaction forces during landing are scarce, the following discussion will explore what is known about other sex differences in landing strategies that may contribute to increased AKSFid as observed in females.

Muscle Activation

A quadriceps dominant activation pattern has been suggested as a possible mechanism for poor sagittal plane knee stabilization leading to increased ACL injuries in females (Boden et al., 2000; Hewett et al., 1996; Huston & Wojtys, 1996; Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001). This contention is primarily supported by observations of females landing in small knee flexion angles during videotape analyses of ACL injuries (Boden et al., 2000), activating the quadriceps prior to the hamstrings in response to anterior translation of the tibia during weight-bearing (Huston & Wojtys, 1996), and when compared to males, exhibiting decreased isokinetic hamstrings torque (Hewett et al., 1996), greater net knee extension moments during landing (Hewett et al., 1996), and greater quadriceps and lesser hamstrings activation throughout the stance phase of the side-step and cross-over cuts (Malinzak et al., 2001). However, none of these studies examined muscle activation strategies and knee flexion angle in combination. Due to the fact that little research has examined this phenomenon via muscle activation patterns during landings, it is not clear if a "quadriceps dominant" muscle activation pattern exists in females during landing. Hence, in light of the possibility of a quadriceps dominant pattern as an explanation for the sex differences observed in AKSFid, the following section will examine what is known about sex differences in muscle activation patterns during landing.

Pre-Landing

Activation of the medial hamstrings has been observed to occur earlier in females than in males prior to the initial contact (173 ± 54 vs. 113 ± 46ms, P = .03) of a single-leg landing, while this same investigation reported no sex differences in the timing of the lateral hamstrings, quadriceps, or gastrocnemius muscles in preparation for landing (Cowling & Steele, 2001). High standard deviations seen in these variables and the relatively small sample size (7 males and 11 females), may have limited the statistical power necessary to detect other sex differences. A close examination of the data shows that the rectus femoris (99 ± 41 vs. 65 ± 30ms, P = .08), vastus lateralis (120 ± 46 vs. 93 ± 29ms, P = .18), and lateral hamstrings (173 ± 54 vs. 115 ± 65ms, P = .06) also activated earlier in females than in males prior to initial contact. These trends also appear to hold in onset times calculated prior to peak AKSF [rectus femoris (128 ± 41 vs. 99 ± 39ms, P = .17), vastus lateralis (153 ± 61 vs. 129 ± 26ms, P = .35), lateral hamstrings (199 ± 48 vs. 151 ± 71ms, P = .10), and medial hamstrings (194 ± 37 vs. 158 ± 48ms, P = .09)] (Cowling & Steele, 2001). Other work reported no sex differences in the activation amplitudes of the quadriceps or hamstrings in preparation for the initial contact of landings, however only six males and eight females were studied and no significance values were reported for the analysis of muscle activation amplitudes prior to landing (Fagenbaum & Darling, 2003). Because of the limited studies and sample sizes to date, it is difficult to draw firm conclusions regarding sex-dependent muscle pre-activation strategies in preparation for landing. Further research is necessary to determine whether males and females differ in pre-activation onset timing and amplitude of the muscles acting on the knee in the sagittal plane, and if differences exist, to what extent they influence AKSFid.

Post-Landing

Little to no sex differences have been reported in post landing onset times or activation amplitudes of the quadriceps, hamstrings, or gastrocnemius muscles (Cowling & Steele, 2001; Fagenbaum & Darling, 2003; Rozzi et al., 1999). Specifically, while females have been reported to exhibit greater peak amplitude and area of the first lateral hamstrings contraction than males following a single-leg landing (Rozzi et al., 1999), similar sex differences have not been observed for peak amplitudes (Fagenbaum & Darling, 2003; Rozzi et al., 1999) or timing (initial contact to peak muscle burst) (Cowling & Steele, 2001) of the quadriceps, medial hamstrings, or gastrocnemius contractions. While one study utilized 17 males and 17 females (Rozzi et al., 1999), sample sizes in the other two studies (Cowling & Steele, 2001; Fagenbaum & Darling, 2003) were relatively small and may not have been sufficient to detect sex differences in muscle activation data. As with muscle pre-activation, more research is needed to determine if sex differences in post-landing activation patterns exist, and to what extent they contribute to or reflect the higher ASKFid in females.

Co-Activation

While sex differences in individual muscle activation patterns during landing are few and inconclusive, the investigation of muscle co-activation patterns during landing may more accurately reflect their contributions to sagittal plane knee joint stability. Little to no published research to date has evaluated co-activation strategies during landing. While a recent report revealed no sex differences in hamstrings to quadriceps amplitude ratios in the last 100ms prior to landing, the first 100ms following the initial contact of a landing, or from initial contact of a landing to maximum knee flexion (Croce, Russell, Swartz, & Decoster, 2004), relative activation ratios alone may not be sufficient to examine this issue. For example, Wjotys et al. (2002) used a simulated weight-bearing model to evaluate the effectiveness of maximum co-contraction of the quadriceps and hamstrings to resist anterior tibia translation during the application of an anterior directed load of 20% body weight on the tibia (i.e. an externally applied AKSF) (Wojtys, Ashton-Miller, & Huston, 2002). Interestingly, males with above average anterior knee laxity (>6mm) were able to generate a larger percentage increase in shear stiffness from rest to maximum co-contraction than females with above average anterior knee laxity (>6mm) (379% vs. 212%, $P = .003$) (Wojtys et al., 2002). This resulted in less anterior tibia translation in males compared to females (2.2 vs. 3.1 mm, $P = .001$) (Wojtys et al., 2002). It appears that relative muscle activations alone are not sufficient to examine this issue. Therefore, when evaluating the effectiveness of thigh muscle co-contraction in controlling AKSFid at a given knee flexion angle, future investigations may benefit from examining both the absolute magnitude of the contraction and the relative balance of quadriceps to hamstring activation.

Anatomical Characteristics

While research to date examining sex differences in muscle activation patterns during landings has yet to explain the higher AKSFid seen in females, sex differences in intrinsic anatomical structure of the lower extremity may interact with the neuromuscular system to control AKSFid. Not only have anatomical characteristics been cited in recent consensus statements as risk factors for ACL injury (Griffin et al., 2000; McClay Davis I, 2003; McClay-Davis & Ireland, 2001), but sex differences in a variety of lower extremity anatomical characteristics exist (Blackburn, Riemann et al., 2004; Cornbleet & Woosley, 1996; Guerra, Arnold, & Gajdosik, 1994; Hahn & Foldspang, 1997; Harner et al., 1994; Huston & Wojtys, 1996; Moul, 1998; Ranovic, 1990; Rosene & Fogarty, 1999; Rozzi et al., 1999; Trimble et al., 2002; Woodland & Francis, 1992). Specific to the sagittal plane, females have been observed to have significantly greater hamstrings extensibility (Blackburn, Riemann et al., 2004) and anterior knee laxity (Huston & Wojtys, 1996; Rosene & Fogarty, 1999; Rozzi et al., 1999; Trimble et al., 2002) than males. Prospective and retrospective investigations indicate that hamstrings extensibility (Boden et al., 2000; Nicholas, 1970) and anterior knee laxity (Woodford-Rogers, Cypher, & Denegar, 1994) may be related to ligamentous injury at the knee. The following discussion will focus on how these two anatomical characteristics may interact with the neuromuscular system to contribute to the sagittal plane knee stabilization measure of AKSFid.

Hamstrings Extensibility

Conventional methods of injury prevention suggest that increased flexibility or extensibility of a muscle may decrease the likelihood of a muscle strain, since less force is generated within the muscle-tendon unit (Cross & Worrell, 1999). However, hamstrings extensibility may not be protective of knee ligament injuries (Boden et al., 2000; Nicholas, 1970). In a prospective investigation of flexibility in football players, 39% of those who were able to touch their palms to the floor with their knees straight (an indirect measure of hamstrings extensibility) sustained a ligament rupture within five years, while only two of 50 who were not able to place their palms on the floor sustained a ligament rupture (Nicholas, 1970). Further, retrospective data indicate that greater hamstrings flexibility was present in a higher percentage of ACL-injured limbs (52%) than control limbs (21%) (Boden et al., 2000). These data suggest that those with increased hamstrings extensibility may not be able to stabilize their knee as effectively, as indicated by increased rates of ligamentous injury.

Increased risk of ligamentous injury due to increased hamstrings extensibility may be explained by an inability of the hamstrings muscle to generate force sufficient to control the development of AKSFid. Hamstrings extensibility has been observed to predict a portion of the variance (15%) in active hamstrings muscle stiffness, with greater extensibility predicting lesser active stiffness (Blackburn, Padua et al., 2004). This relationship between muscle extensibility and stiffness may be explained by longer muscle fiber lengths, smaller pennation angles, and decreased muscle thickness in females compared to males even when these variables are normalized to limb length

(Kanehisa, Muraoka, Kawakami, & Fukunaga, 2003; Kubo et al., 2003). Hence, females may produce less muscle-tendon force at a given joint angle secondary to an altered length-tension relationship (Zajac, 1989). This altered length-tension relationship in females with increased hamstrings extensibility may reduce the passive stiffness contribution to the overall force production capability of the muscle-tendon unit (Robertson et al., 2004), thereby reducing the ability of the hamstrings to produce the force necessary to control AKSFid at given joint angle. While more work is needed to establish these theoretical links, the collective findings to date suggest that at a given knee flexion angle, females with greater hamstrings extensibility may not be able to generate as much force in the hamstrings to resist AKSFid during the deceleration phase of landing. Should this theory be proven in future investigations, this would suggest that under similar hamstrings to quadriceps activation ratios, females with increased hamstring extensibility may not be as effective at stabilizing the knee in the sagittal plane, potentially resulting in increased AKSFid.

Anterior Knee Laxity

Anterior knee laxity has also been suggested as a risk factor for ACL injury (Uhorchak, Scoville, Williams, St. Pierre, & Taylor, 2003; Woodford-Rogers et al., 1994), and has been observed to be greater in females than males (Huston & Wojtys, 1996; Rosene & Fogarty, 1999; Rozzi et al., 1999; Trimble et al., 2002; Uhorchak et al., 2003; Woodford-Rogers et al., 1994). Anterior knee laxity has been observed to be greater in ACL injured than uninjured Army cadets, and those with anterior knee laxity greater than one standard deviation above the mean were 2.7 times more likely than others to sustain ACL injury (Uhorchak et al., 2003). In combination, anterior knee laxity and navicular drop have been observed to predict the ACL injury status of 87.5% of athletes and to predict 60% of the variance between injured and uninjured female athletes, with increased values being related to an increased risk of ACL injury (Woodford-Rogers et al., 1994). Despite these results, the influence of anterior knee laxity on measures of sagittal plane knee joint stabilization during landing has received little attention to date.

During single-leg weight-bearing perturbations described to mimic the side-step and cross-over cut (Schmitz, Shultz, Kulas, Windley, & Perrin, 2004), females with above average knee laxity (>7mm) displayed increased pre-activation amplitudes and a greater increase in the reflex activation amplitude of the lateral hamstrings than females with below average knee laxity (<5mm) (Shultz, Carcia et al., 2004). While these findings suggest a muscle activation pattern protective of the development of AKSFid during a perturbation, the author is unaware of any investigations examining these effects during landing. Similar parallels, however, may be made in the ACL-deficient (ACLd) population.

ACLd is associated with greater anterior knee laxity in the involved knee as compared to the uninvolved knee (Gauffin & Tropp, 1992) and the knees of healthy controls (Wojtys & Huston, 1994). When compared to controls, a larger percentage (50% vs. 38%) of persons who are ACLd respond to an applied AKSF with the hamstrings prior to the quadriceps or gastrocnemius (Wojtys & Huston, 1994). Other work has shown that persons who are ACLd synchronize activation of the hamstrings more closely with the time of initial contact and peak AKSFid during a single-leg landing when compared with controls (Steele & Brown, 1999). Further, the quadriceps of the involved limb exhibit less activity than the quadriceps in the uninvolved limb (Gauffin & Tropp, 1992) and in the limb of healthy controls (Swanik, Lephart, Giraldo, DeMont, & Fu, 1999) during single-leg landings. Collectively, these data suggest that muscle activation patterns in females with high knee laxity and in persons who are ACLd appear to be protective of AKSFid through modulation of quadriceps and hamstrings activation patterns. Further studies are necessary to confirm if this protective mechanism exists during landing in healthy females with high knee laxity.

Summary

Little research to date has examined sex differences in individual muscle activation patterns prior to and following landing with sufficient sample sizes, and therefore, it is difficult to draw conclusions as to how these patterns may influence AKSFid. However, other data suggest that females may not be able to control externally applied AKSF with the same efficiency as males during maximum co-contraction of the quadriceps and hamstrings. An interaction between sex-dependent anatomical characteristics and the efficiency of the neuromuscular system to contribute to controlling AKSF may explain this phenomenon. Specifically, females with greater hamstrings extensibility may have a reduced ability to produce muscle-tendon force in the hamstrings at a given joint angle, possibly limiting their ability to resist the AKSFid that occurs during landing. Females with greater knee laxity, on the other hand, may exhibit quadriceps and hamstrings firing patterns that may protect the knee joint from greater

AKSFid during a landing. The clinical implications of these findings are that while females with greater anterior knee laxity may effectively stabilize the knee joint in the sagittal plane through increased hamstrings activation, females who also have greater hamstrings extensibility may not be able to generate enough hamstrings force to effectively control AKSFid. Further research is necessary to confirm these theoretical links and to evaluate the interactions of anatomical characteristics and neuromuscular activation patterns as they contribute to the control of AKSFid during landings in females.

Summary

Female athletes injure the ACL of the knee during sport more frequently than male athletes, and a large percentage of these injuries occur during landing. Inadequate sagittal plane knee joint stabilization may put the ACL at risk for injury during landings, and AKSF represents a measure of sagittal plane knee joint stabilization that has been implicated as a risk factor for ACL injury due to its relationship with anterior translation of the tibia on the femur, ACL tension, and ACL strain. As estimated through inverse dynamics, AKSFid represents the sum of all the biomechanical and neuromuscular forces acting on the knee in the sagittal plane. Data suggest that females experience higher AKSFid than males during landing, however, the factors or events that predispose females to greater AKSFid and ACL injury have yet to be elucidated. Sufficient evidence is not yet available to determine if sex differences in the biomechanical and neuromuscular events of landing alone may explain sex discrepancies in AKSFid.

Females do, however, appear to have a decreased ability to resist externally applied AKSFid than males during similar levels of co-contraction of the quadriceps and hamstrings. While it is yet unknown how anatomical characteristics intrinsic to females may interact with or cause this set of circumstances that leads to high AKSFid, hamstrings extensibility and anterior knee laxity may interact with the neuromuscular system in a manner that may, in part, explain this phenomenon.

CHAPTER III

METHODS

Design

Multiple linear regression analyses were used to evaluate the ability of hamstrings muscle extensibility (HS), anterior knee laxity (KT), and hamstrings and quadriceps muscle activation amplitudes to predict AKSFid during single-leg landings in healthy, recreationally active females. Clinical measures of HS and KT on the preferred leg were averaged over three trials for data analysis. The landing task was performed with the participant balancing on the preferred leg on a 30cm platform and dropping onto the same leg on a force plate with the center positioned 30% of the participant's height from the front edge of the platform. Kinetic and electromyographic data were collected over five trials and later averaged for data analysis. HS, KT, and normalized hamstrings and quadriceps pre-landing activation amplitudes (H_{pre} & Q_{pre} respectively) were used to predict initial AKSF (iAKSF), while HS, KT, H_{pre}, Q_{pre} and normalized hamstrings and quadriceps post-landing activation amplitudes (H_{post} & Q_{post} respectively), were used to predict rate of AKSF (rAKSF) and peak AKSF (pAKSF).

Participants

Forty-five healthy, recreationally active females between the ages of 18 and 30 were recruited from the University of North Carolina at Greensboro and the surrounding community. All participants were recruited based on the following inclusion criteria: no injuries to the lower extremities in the past six months; no history of ligamentous rupture in the feet, ankles, knees, or hips; no history of surgery to the lower extremities; and current participation in recreational exercise for a minimum of 90 minutes per week. Participants with a history of injury to the lower extremities in the past six months, ligamentous injury, or surgery were excluded due to possible influence on the on the predictor variables. Only females were included in this study in an effort to eliminate other potential sex-confounding factors on the relationships examined. All data were collected during the first eight days of the menstrual cycle due to evidence of hormonal effects on KT throughout the rest of the menstrual cycle (Shultz, Kirk, Johnson, Sander, & Perrin, 2004). Participants were instructed to avoid exercise the day of and the day prior to data collection to remove the potential of exercise induced effects on anterior knee laxity or hamstrings extensibility. No participants were excluded from this study based on race or ethnicity, and participants were instructed that they may withdraw from the study at any time. No participants elected to withdraw. Previously collected data during single-leg drop landings from 45cm (Tables 1 & 2) indicate that, to achieve a statistical power of .80 with an a priori alpha level set at .05, 31 participants were needed to establish statistically significant stepwise multiple linear regressions for the prediction of iAKSF and rAKSF from HS and KT. A conservative estimate of 14 participants was added to account for the reduction in power expected by adding H_{pre}, Q_{pre}, H_{post}, and Q_{post} to the prediction equations. All power and sample size estimates were calculated using SPSS Sample Power Version 2.0 (SPSS Inc.; Chicago, IL). The correlational relationships between the variables and the results of the regression power analyses can be seen in Tables 1 and 2, respectively. The neuromuscular measures was not accounted for in the pilot investigation.

Instrumentation

HS was measured with a standard goniometer modified with an extension piece on the movable arm to better approximate the line of the thigh. KT was measured with a KT 2000 Knee Arthrometer (Medmetric Corp, San Diego, CA) with a level secured to the superior surface to ensure straight sagittal plane movement. Kinematic data for the pelvis and both thighs, shanks, and feet were sampled at 140 Hz using an electromagnetic tracking system (Ascension Technology; Burlington, VT) and Motion Monitor software (Innovative Sports Training; Chicago, IL) during the landing task. A type 4060 non-conducting force plate (Bertec Corporation; Columbus, OH) and the Motion Monitor software sampled ground reaction forces at 1000Hz during the landing task. Surface electromyographic (sEMG) signals were collected using a 16 channel Myopac telemetric system (Run Technologies; Mission Viejo, CA) with 10mm bipolar Ag-AgCl surface electrodes (Medicotest Blue Sensor #N-00-S; Ambu Products, Germany) placed halfway between the motor point and the distal tendon of the vastus lateralis and medial and lateral hamstrings (Rainoldi, Melchiorri, & Caruso, 2004) during maximum voluntary isometric contractions (MVICs) of the quadriceps and hamstrings and during the landing task. The Myopac unit has an amplification of 1mV/V with a frequency bandwidth of 10 to 1000Hz, a common mode rejection ratio of 90dB min at

Table 1. Correlational Relationships from Pilot Testing

Bivariate Pearson Correlations

	KT	HS	iAKSF	pAKSF	rAKSF
KT	1				
HS	0.60	1			
	0.02*				
iAKSF	-0.43	-0.47	1		
	0.11	0.08			
pAKSF	-0.13	-0.04	-0.21	1	
	0.66	0.89	0.46		
rAKSF	-0.34	0.10	0.07	0.36	1
	0.22	0.72	0.80	0.18	

*Value is Significant at alpha = .05

Table 2. Regression Results from Pilot Testing

Regressions & Effects

Reg Model	DV	PV1	PV2	Sig	R^2	Adj R^2	f^2	n
Stepwise	iAKSF	HS	KT	0.17	0.259	0.136	0.35	31
Stepwise	rAKSF	KT	HS	0.17	0.254	0.129	0.34	31

DV - Dependent Variable

PV - Predictor Variable

Sig - P-value of the sums of squares test for significance of the regression

R^2 - Regression coefficient

Adj R^2 - Adjusted regression coefficient

$f^2 = R^2 / 1 - R^2$ or the effect size (small <0.15; medium < 0.35; large > 0.35) (Cohen & Cohen, 1988)

n - sample size needed to detect a significant regression with a power = 0.80 and alpha = 0.05

60Hz, an input resistance of 1 MΩ, and an internal sampling rate of 8 KHz. A Biodex System 3 isokinetic dynamometer (Biodex Medical Systems Inc.; Shirley, NY) was used to position the participant at a fixed knee flexion angle of 30° during the MVIC trials.

Procedures

Demographic Measures

Upon arrival to the Applied Neuromechanics Research Laboratory each participant read, signed, and was provided a copy of a written informed consent approved by the University of North Carolina at Greensboro's Institutional Review Board for the Protection of Human Participants. The complete consent form can be found in Appendix A. The participant then filled out an activity and injury questionnaire that detailed their lower extremity injury history, previous experience with landing activities, and date on which they began their most recent menstrual cycle. This questionnaire can be found in Appendix B. The participants' height and weight were then measured with a standard digital scale and a measuring tape mounted on the wall. Thirty percent of the height of the participant was calculated for positioning of the landing platform. These data were recorded in writing. The complete data collection sheet used for this study can be found in Appendix C.

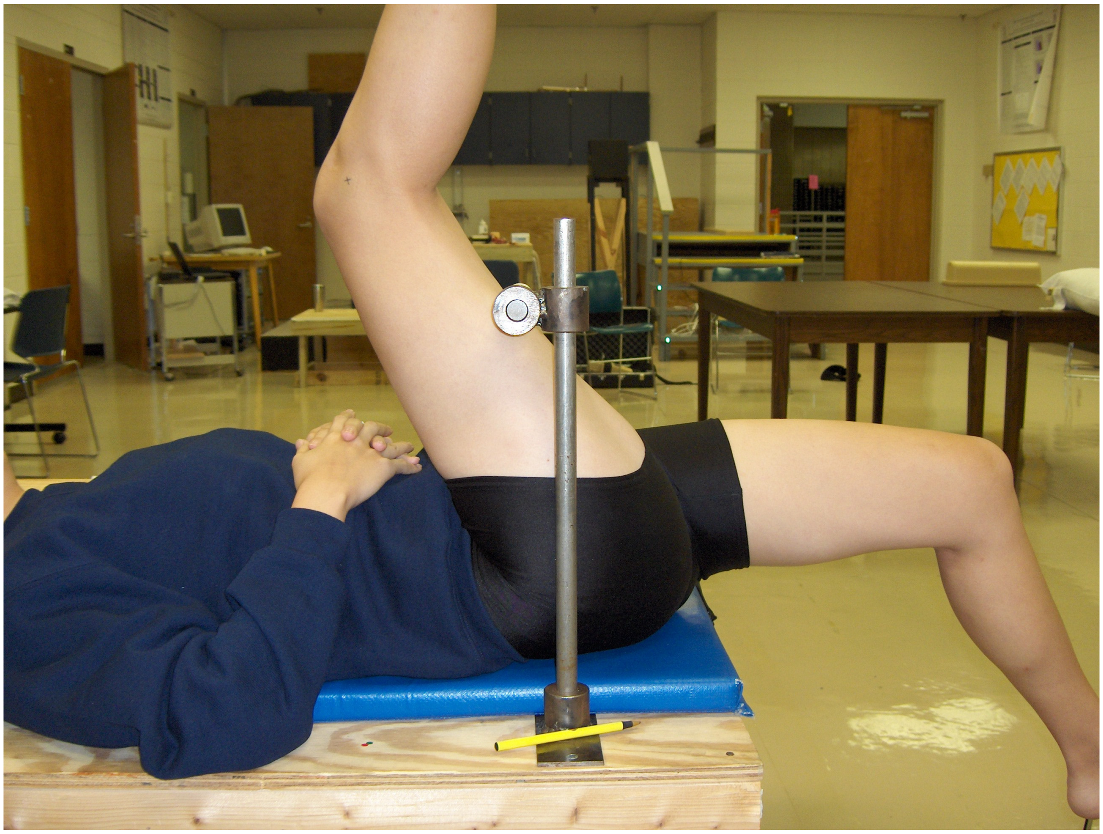
Clinical Anatomical Measures

HS and KT were then measured by the examiner on one limb and then on the contralateral limb. The first limb measured was counterbalanced across participants.

Hamstrings Extensibility

Hamstrings extensibility (HS) was measured by a technique modified from Blackburn et al. (Blackburn, Riemann et al., 2004) The most prominent part of the lateral malleolus was marked with indelible ink. The participant was then positioned in supine on a table with the hip of the test limb flexed to 120° and the contralateral hip placed at zero degrees with the foot supported. A bar mounted on a steel frame affixed to the table was used as a tactile cue for the participant to control the hip flexion angle. The bar was placed at a height that was halfway between the knee and hip joints when the hip was in 120° of flexion and was adjusted to the steel frame so that a moment greater than 5Nm would rotate the bar. The participant was instructed to straighten their knee as far as possible without rotating the bar and then hold that position for approximately two seconds (Figure 2). Previous investigations have positioned the participants in 90° of hip flexion (Blackburn, Padua et al., 2004; Nyland, Caborn, Shapiro, Johnson, & Fang, 1999; Swanik, Lephart, Swanik, Stone, & Fu, 2004), however several females were able to reach full knee extension during pilot testing. Hence, the participants were positioned in 120° to ensure that the entire range of variability was appreciated for this measure. Five conditioning trials were completed to reduce the effects of task learning and/or increases in tissue extensibility. During the five conditioning trials the anterior-posterior center of the knee joint and the most prominent aspect of the lateral malleolus were marked with indelible ink when the knee was extended as far as the participant could tolerate. Then, three test trials measured the angle formed by a line from the greater trochanter to the mark on the knee joint and from the mark on the knee joint to the mark on the lateral

Figure 2. Hamstrings Extensibility

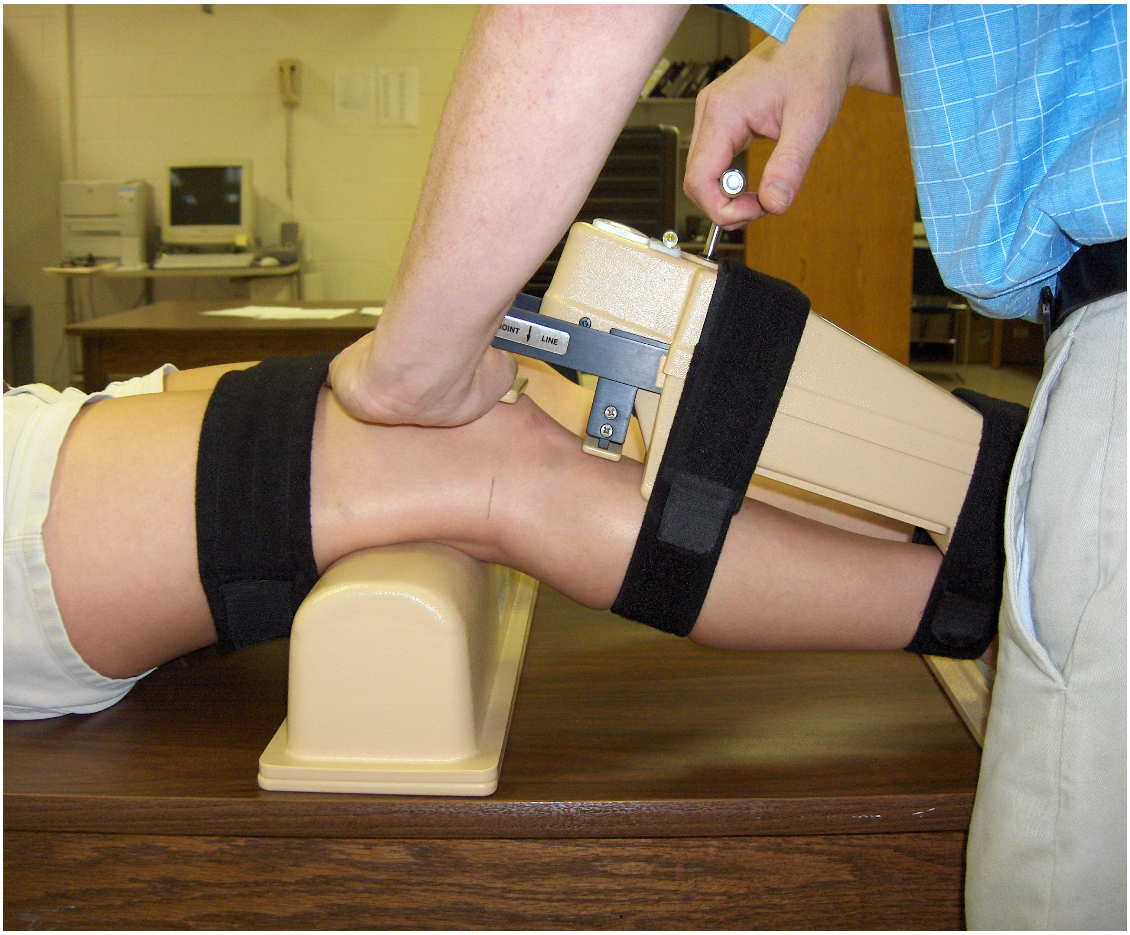


malleolus to the nearest degree with a universal goniometer. A larger angle indicated a more extensible hamstrings muscle. Any trials in which the horizontal bar was rotated were discarded and repeated. The examiner previously demonstrated excellent day to day reliability ($ICC_{2,3}$ = .98, SEM = 1.9°) for this measure.

Anterior Knee Laxity

KT was measured with the participant positioned supine on a table with the head on a pillow and the knee in 25° to 30° of flexion (confirmed by a standard goniometer) over a four inch bolster. The participant's lateral joint line of the knee was then marked with indelible ink. The participant was instructed to "relax their leg muscles" while the tester placed a strap around the thighs of the participant to align the tibial crests to the vertically (i.e. perpendicular to the horizontal). The KT 2000 was then applied using the manufacturer's guidelines with the lateral joint line of the knee in line with the mark on the side of the KT 2000, the proximal strap snug over the belly of the calf, and the distal strap snug over the distal shank. The height of the KT was adjusted to align the displacement dial, and the medial-lateral rotation of the device was aligned such that the bubble level was centered. Prior to measurement of anterior knee laxity the investigator zeroed the KT 2000 by imposing a 45N posterior force on the tibia relative to the femur and releasing this force. Each time the investigator then adjusted the displacement dial to align with zero. Once the posterior force was released and the dial returned to zero, the investigator measured the anterior movement of the tibia on the femur in the sagittal plane to the nearest half of a millimeter when 133N of anteriorly directed force was applied to the posterior tibia (Figure 3). Three trials were completed, and prior to each

Figure 3. Anterior Knee Laxity



trial the investigator repeated the above zeroing procedures to ensure maximum measurement consistency. The examiner previously demonstrated excellent day to day reliability ($ICC_{2,3}$ = .97, SEM = .4mm) for this measure.

Landing Instruction and Practice

The front edge of the landing platform was positioned 30% of the height of the participant behind the center of the force plate. The single-leg landings were instructed, demonstrated, and then practiced by each participant. They were instructed to "place their hands on their hips, balance on one leg, lean forward, drop off the platform, and land on the same leg in the center of the force plate." The examiner also explained that the participant should "drop, not jump," off the box onto the force plate and hold their balance for a minimum of one second while keeping their hands on their hips throughout the trial. The participant then practiced the task three times. The preferred leg was defined as the leg on which the participant landed in two of the first three practice trials. The landing was then demonstrated by the examiner using the designated preferred leg. The participant then practiced the landing until the examiner determined that the participant was able to perform the landing task consistently as instructed. All landings were performed in bare feet. The positioning of the participant on the landing platform can be visualized in Figure 4.

Surface Electrode Preparation

The areas over the muscle bellies of the vastus lateralis, medial and lateral hamstrings, and anterior patella of the preferred leg were shaved and scrubbed with alcohol pads to remove any substances that might prevent optimal surface contact or

Figure 4. Drop Landing Position

conductivity. Pairs of surface electrodes were placed halfway between the motor point and the distal tendon of the vastus lateralis and medial and lateral hamstrings (Rainoldi et al., 2004) with a center to center distance of 2.5cm (Shultz et al., 2005). A reference electrode was placed over the ipsilateral patella. The positioning of these electrodes can be visualized in Figure 5.

Maximum Voluntary Isometric Contractions

The participant was then secured in the Biodex with the knee fixed in 30° of flexion (Figure 6). The axis of the movement arm of the dynamometer was aligned with the sagittal axis of the knee joint and the resistance pad of the dynamometer was secured directly proximal and anterior to the superior portion of the medial malleolus. The participant’s torso, hips, and thigh of the preferred leg were secured with the Velcro straps on the Biodex. Prior to testing five seconds of baseline sEMG data were collected and the oscilloscope was visualized to detect any extraneous noise and cross-talk. The participants were then instructed to cross their arms over their chest and kick (knee extension) or pull (knee flexion) against the fixed resistance of the dynamometer as hard as possible for five seconds. One practice trial was performed prior to the first trial, followed by one minute of rest, followed by three, five second maximal volitional isometric contractions (MVICs) of the quadriceps (knee extension) and hamstrings (knee flexion) interspaced with one minute rest intervals. The oscilloscope was visualized during each trial to ensure high signal quality and to check for cross-talk. The sEMG electrodes recorded muscle activation throughout the MVIC trials. The sEMG data were

Figure 5. Surface Electrode Positions

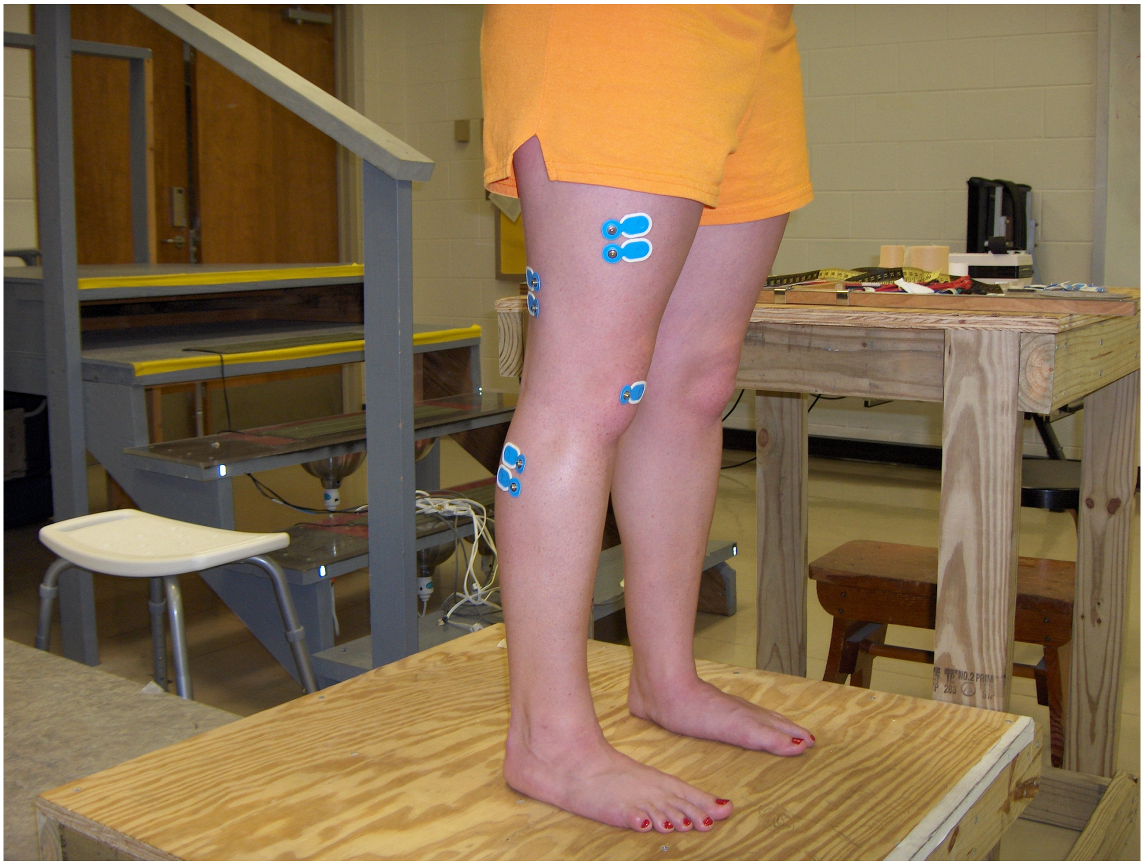
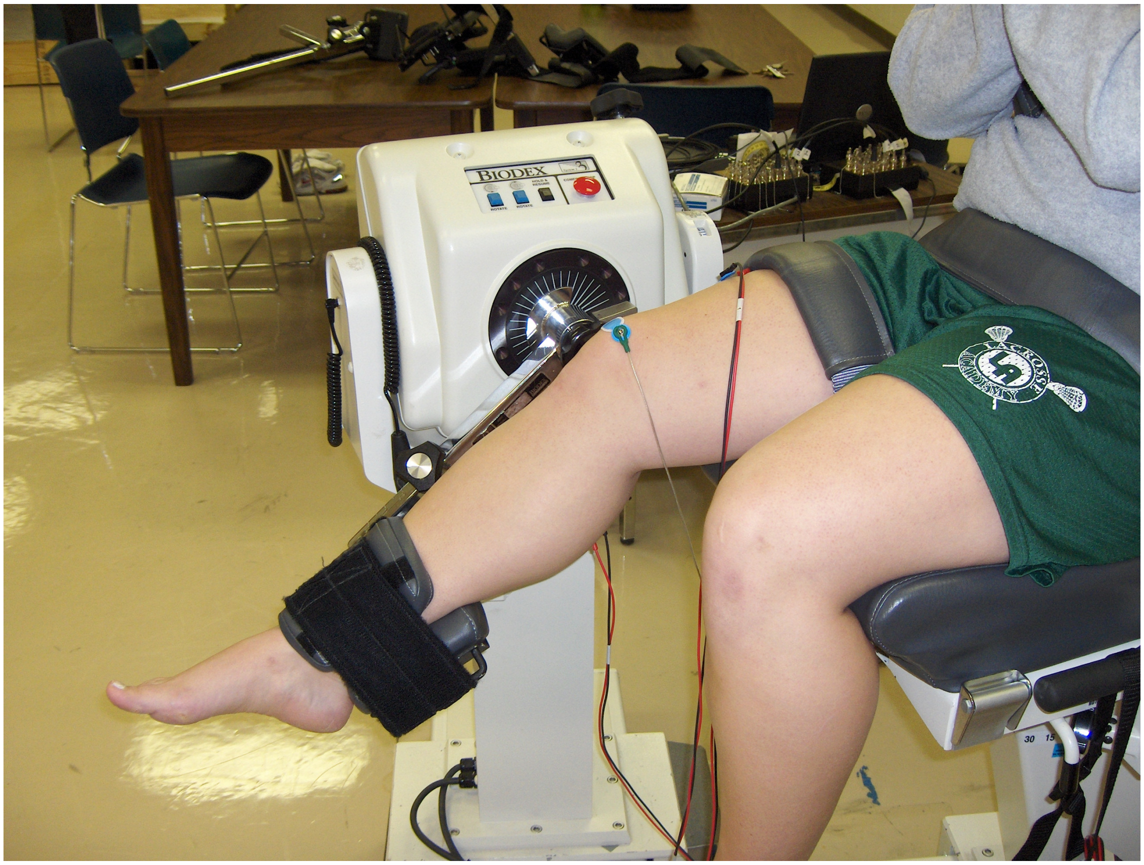


Figure 6. MVIC Position



collected and stored in Datapak and later used to normalize the sEMG data collected during the landing trials.

Landing Trials

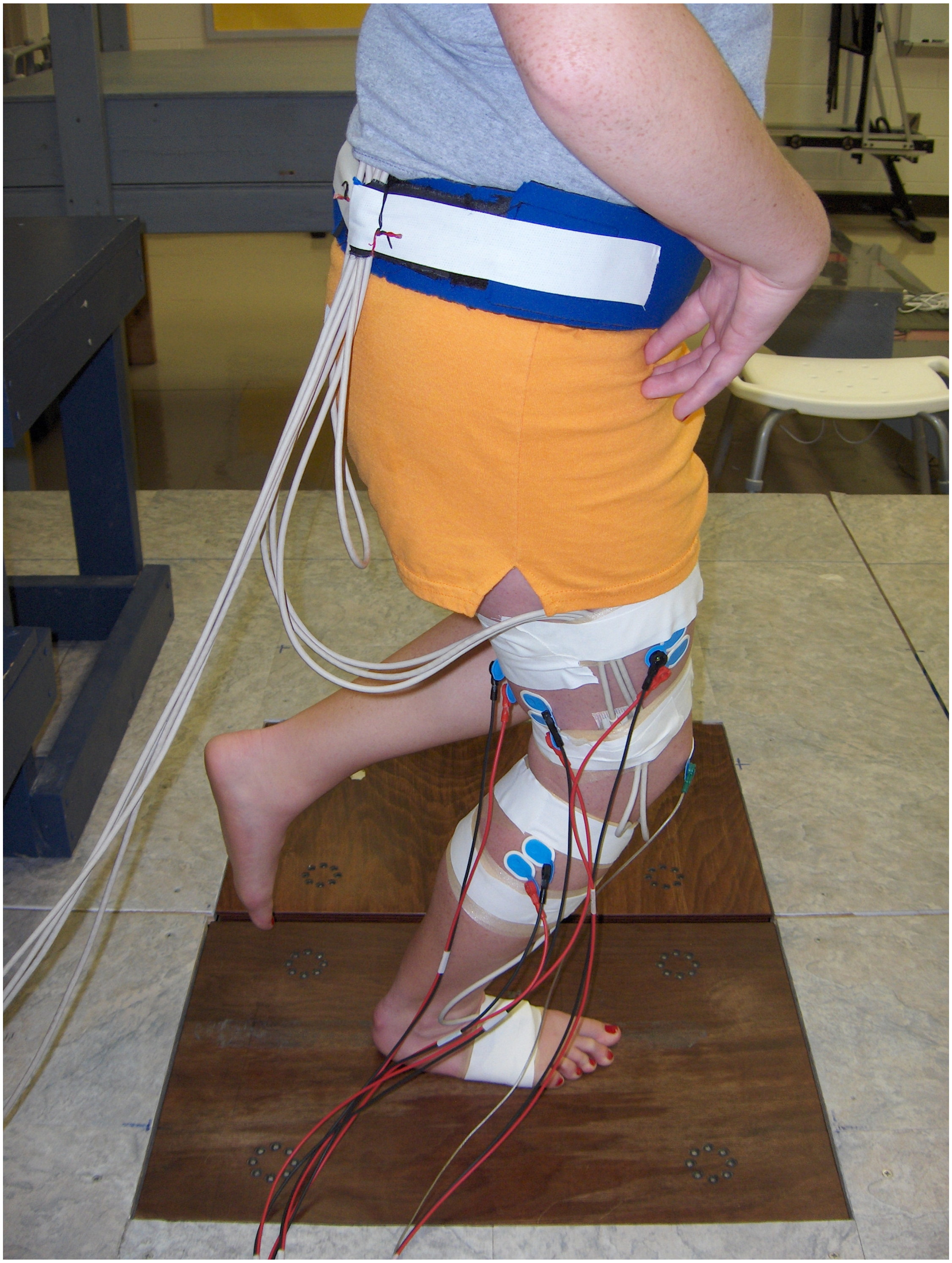
Six-degree-of-freedom electromagnetic sensors were then secured with double-sided tape and/or Velcro straps and athletic tape to the sacrum, and the lateral aspect of the femur, mid-shaft of the anterior-medial tibia, and head of the third metatarsal of the preferred leg. Joint centers of each participant were estimated as the midpoints between the medial and lateral femoral condyles for the knee, the medial and lateral ankle joint lines just distal to the medial and lateral malleoli for the ankle, and from a series of thigh positions relative to the sacrum for the hip (Leardini et al., 1999; Madigan & Pidcoe, 2003). Following the digitization procedure, and prior to the landing trials, each participant was instructed to sit completely still on a stool with their arms at their side for one second. During this time baseline sEMG data for the quadriceps and hamstrings were collected to later remove any baseline noise from the sEMG data created by the electromagnetic tracking system. Each participant then performed five acceptable trials of the single-leg landings on the preferred leg while kinematic, kinetic, and electromyographic data were simultaneously recorded. Initial ground contact of each landing was defined as the first frame of vertical ground reaction force data equal to or greater than 40N. Vertical ground reaction force data were monitored synchronously by the Motion Monitor and Datapak, and initial ground contact triggered the simultaneous recording of 500ms of data prior to initial ground contact and 500ms of data following initial ground contact. Acceptable landing trials were defined as those in which the participant dropped, not hopped or lowered, off the box [determined by sacral sensor height remaining within plus or minus 1.5cm of the first frame of data (500ms prior to landing) during the time on the landing platform]; landed on the preferred leg in the center of the force plate and held their balance for a minimum of one second; did not hop upon landing; and kept their hands on their hips throughout the landing and for a minimum of one second following the landing. AKSFs normalized to body weight (BW) and body weights per second (BW/sec) have previously been demonstrated in our laboratory to exhibit high day to day reliability for iAKSF ($ICC_{2,5}$ = .93, SEM = .01BW), rAKSF ($ICC_{2,5}$ = .91, SEM = 1.55BW/sec), and pAKSF ($ICC_{2,5}$ = .93, SEM = .04BW) during a single-leg drop landing from 30cm using the same landing instructions and kinematic and kinetic methods (Kulas, unpublished data). The position of the participant during the single-leg landing can be visualized in Figure 7.

Data Processing

Demographic and Clinical Measures

The means and standard deviations of the participants' age, height, weight, number of exercise sessions per week, and duration of exercise sessions were calculated to report participant demographics. The average of the three trials of HS and KT were calculated for each participant for the preferred and contralateral leg. The preferred leg was used for all statistical analyses.

Figure 7. Single-leg Landing



Kinetic Measures

All kinematic data of the thigh referenced to the shank were estimated using a three-dimensional, link-segment model with Euler angles. Kinematic data were interpolated to align with kinetic data, and were low pass filtered at 12Hz using a 4^{th} order, zero-lag digital Butterworth filter, while kinetic data were low pass filtered at 60 Hz using a 4^{th} order, zero-lag digital Butterworth filter. Kinetic data of the thigh relative to the shank reference frame were estimated with inverse dynamics solutions using anthropometric data estimated from the mass and height of the participant, and ground reaction forces and kinematic data estimated by the Motion Monitor. All kinetic data were then exported into a spreadsheet for calculation of AKSFs. A posterior shear force of the thigh relative to the shank reference frame was defined as a positive AKSF (Moglo & Shirazi-Adl, 2003). The iAKSF was defined as the amount of AKSF at the time of initial ground contact (40N of vertical ground reaction force). The pAKSF was defined as the greatest AKSF value following initial ground contact. Both iAKSF and pAKSF were normalized to units of body weight (BW) by dividing the force by the weight of the participant in Newtons. The time from initial ground contact to pAKSF was calculated for each trial to calculate rAKSF. The rAKSF was defined as the pAKSF minus the iAKSF, with that difference divided by the time from initial ground contact to pAKSF in seconds. The rAKSF was normalized to units of body weights per second (BW/s). Means and standard deviations of iAKSF, pAKSF, and rAKSF from the five landing trials were calculated and used for all statistical analyses.

Surface Electromyographic Measures

The middle three seconds of the five second MVIC trials for the quadriceps and medial and lateral hamstrings muscles were digitally processed using a root mean square (RMS) algorithm with a 100ms time constant (Shultz et al., 2005). The peak RMS amplitude during the middle three seconds was identified and averaged over the three trials for each muscle. The mean peak RMS amplitude was then used to normalize the muscle activation amplitudes during the landings, expressed as a percentage of their MVIC (%MVIC).

Data for sEMG signals from the landings were low pass filtered at 250Hz and high pass filtered at 15Hz using a 4th order, zero-lag, digital Butterworth filter (Steele & Brown, 1999). The filtered signal was then processed with an RMS algorithm with a 15ms time constant (Swanik et al., 1999), and passive demeaning was performed to subtract out the baseline noise in the sEMG signal for each muscle, estimated from the middle .80sec of the resting baseline sEMG trial acquired prior to the landings. The sEMG signals acquired from the vastus lateralis and medial and lateral hamstrings were ensemble averaged across the five trials to obtain one representative signal for each muscle from which to perform subsequent analyses. Using the averaged signal, interval event markers were then set for 100ms prior to and following the initial contact of the landing. The mean RMS amplitudes during the last 100ms prior to initial ground contact and the first 100ms following initial ground contact were determined and then exported to a computer database program and normalized to the respective MVIC data. H_{pre} was defined as the grand mean %MVIC amplitude of the medial and lateral hamstrings, and

Q_{pre} was defined as the mean %MVIC amplitude of the vastus lateralis during the last 100ms prior to the initial ground contact of landing (Croce et al., 2004). H_{post} was calculated as the grand mean %MVIC amplitude of the medial and lateral hamstrings, and Q_{post} was defined as the mean %MVIC amplitude of the vastus lateralis during the 100ms immediately after initial ground contact (Croce et al., 2004).

Statistical Analyses

Three separate stepwise linear regressions were used to determine the predictive relationships between the predictor variables and the dependent variables.

1. To test hypothesis 1: HS, KT, H_{pre} and Q_{pre} were used to predict iAKSF.
2. To test hypothesis 2: HS, KT, H_{pre}, Q_{pre}, H_{post}, and Q_{post} were used to predict rAKSF.
3. To test hypothesis 3: HS, KT, H_{pre}, Q_{pre}, H_{post}, and Q_{post} were used to predict pAKSF.

The stepping method criteria specified that of probability of $F \leq .49$ was used for entry, and a probability of $F \geq .50$ was used for removal. An assessment of multicollinearity ensured that each predictor variable contributed uniquely to each regression. A tolerance of less than 0.2 was considered unacceptable (Howell, 2002). Leverage statistics were calculated for all 45 participants to examine the influence of the predictor variables on each separate regression. Participants with leverages greater than $3(p + 1)/n$ where p is the number of predictor variables and n is the number of participants [$3(4 + 1)/45 = .33$ for hypothesis 1; $3(6 + 1)/45 = .47$ for hypotheses 2 and 3] were flagged as possible outliers that had influence on the regressions (Howell, 2002). The standardized Z-residual statistic was calculated for all 45 participants to determine the influence of the dependent variables on each separate regression. Participant's with standardized Z-residuals with an absolute value of greater than two were flagged as possible outliers that had influence on the regressions (Pedhazur, 1997). Cook's D was then calculated for each of the flagged participants to evaluate the overall influence of those participants on the regressions. A histogram of Cook's D was visualized to determine if these participants should be removed from the analyses, and those with a Cook's D that visually deviated from the histogram of any of the three regressions were removed from all three analyses (Howell, 2002).

CHAPTER IV

RESULTS

Descriptive Data

The complete set of raw data for subject demographics, predictor variables, and dependent variables can be found in Appendices D, E, and F, respectively. Forty-five healthy, recreationally active female participants (21.6 ± 3.0yrs; 164.7 ± 6.9cm; 62.7 ± 13.3kg) completed data collection successfully, and no participant took greater than 20 attempts to complete the five acceptable landing trials. No violations of multicollinearity were present, however two participants were excluded from all analyses due to violation of the regression outlier criteria (Appendices G through P). For the remaining 43 participants (21.7 ± 3.0yrs; 164.7 ± 7.0cm; 62.7 ± 12.3kg), Tables 3 and 4 list the means and standard deviations for the predictor and dependent variables respectively. Appendices Q through V and W through Y display the spread in the data for the predictor variables and dependent variables respectively. On average, participants exercised 5.5 ± 3.2 hours per week, with 1.1 ± 1.5 of those hours including landing activities. Thirty-three of the participants self-selected the right leg as the preferred leg for landing, and 31 reported experience with landing activities prior to the six months preceding the study. Participants landed with an average knee flexion angle of 16 ± 6° upon initial contact and reached a peak knee flexion angle of 54 ± 10°, over a time of 206 ± 73ms. Peak vertical

Table 3. Means ± Standard Deviations (Sd) for Anatomical and Neuromuscular Predictor Variables

Predictor Variable	**Mean ± Sd**
Hamstrings Extensibility (deg)	138.6 ± 13.1
Anterior Knee Laxity (mm)	7.0 ± 1.8
Hamstrings Pre-landing Amplitude (%MVIC)	34.9 ± 14.0
Hamstrings Post-landing Amplitude (%MVIC)	47.5 ± 52.0
Quadriceps Pre-landing Amplitude (%MVIC)	50.6 ± 28.0
Quadriceps Post-landing Amplitude (%MVIC)	116.3 ± 85.3

N = 43

Table 4. Means ± Standard Deviations (Sd) for Kinetic Dependent Variables

Dependent Variable	Mean ± Sd
Initial Anterior Knee Shear Force (BW)	-0.13 ± .19
Rate of Anterior Knee Shear Force (BW/s)	11.8 ± 4.0
Peak Anterior Knee Shear Force (BW)	0.87 ± 0.15

N = 43

ground reaction forces averaged 4.9 ± .7BW and occurred within 37 ± 7ms. Anterior knee shear forces peaked within 89 ± 19ms at a knee flexion angle of 42 ± 9°.

Statistical Results

Hypothesis I: Prediction of Initial Anterior Knee Shear Force

Examination of the regression coefficients revealed that, H_{pre} negatively predicted iAKSF, thereby partially supporting hypothesis one. In the stepwise linear regression, H_{pre} entered the model first with the highest zero-order correlation with iAKSF (r = -.389), predicting 13.1% of the variance in iAKSF (AdjR^2, F $change_{(1,41)}$ = 7.318, P = .01). Once H_{pre} was accounted for, KT had the highest partial correlation ($r_{partial}$ = .232) and entered the model next, followed by Q_{pre} ($r_{partial}$ = -.207), which together predicted an additional 4.1%. However, the F change for the second (F $change_{(2,40)}$ = 2.281, P = .134) and third (F $change_{(3,39)}$ = 1.743, P = .194) predictor models were not significant. The regression equation for the single predictor model is as follows:

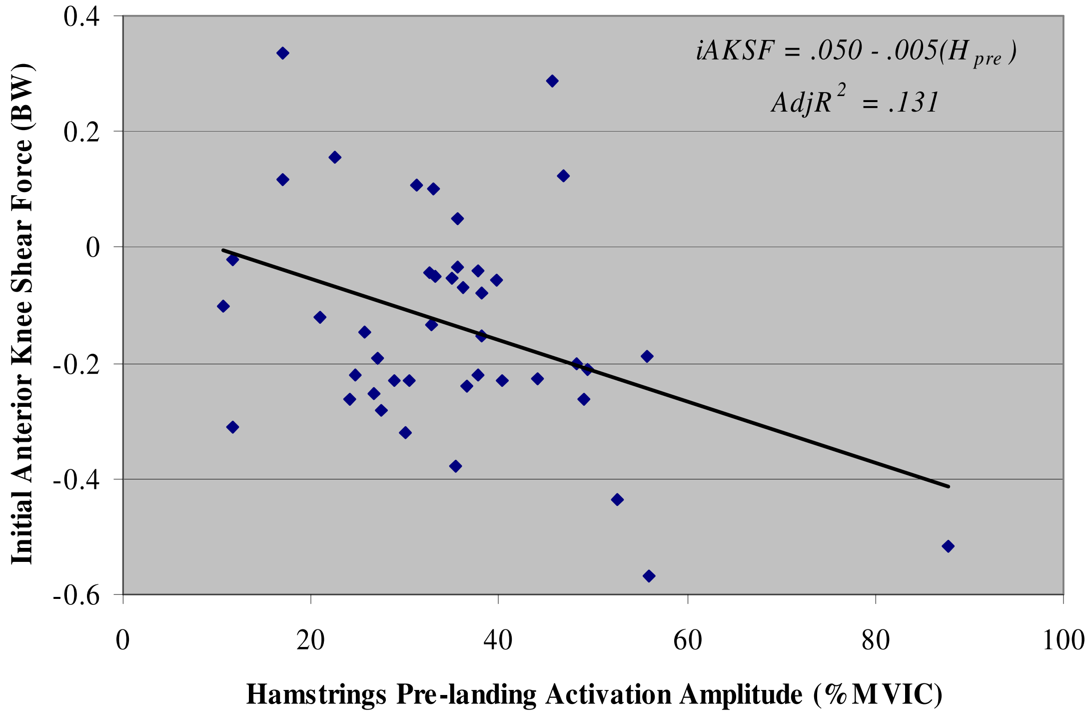
$$iAKSF = .050 - .005(H_{pre})$$

The graphic relationship can be viewed in Figure 8. The full stepwise regression Correlations, Model Summary, ANOVA results, and Coefficient tables can be found in Appendix Z.

Hypothesis II: Prediction of Rate of Anterior Knee Shear Force

Examination of the regression coefficients revealed that, H_{pre} positively predicted rAKSF, thereby rejecting hypothesis two. In the stepwise linear regression, H_{pre} entered the model first with the highest zero-order correlation with rAKSF (r = .412), predicting

Figure 8. Hamstrings Pre-landing Activation Predicted Initial Anterior Knee Shear Force



14.9% of the variance in rAKSF upon landing (AdjR^2, F $change_{(1,41)}$ = 8.380, *P* = .006). Once H_{pre} was accounted for, KT had the highest partial correlation ($r_{partial}$ = -.234) and entered the model next, followed by H_{post} ($r_{partial}$ = .156), which together predicted an additional 2.6%. However, the F change for the second (F $change_{(2,40)}$ = 2.313, *P* = .136) and third (F $change_{(3,39)}$ = .977, *P* = .329) predictor models were not significant. The regression equation for the single predictor model is as follows:

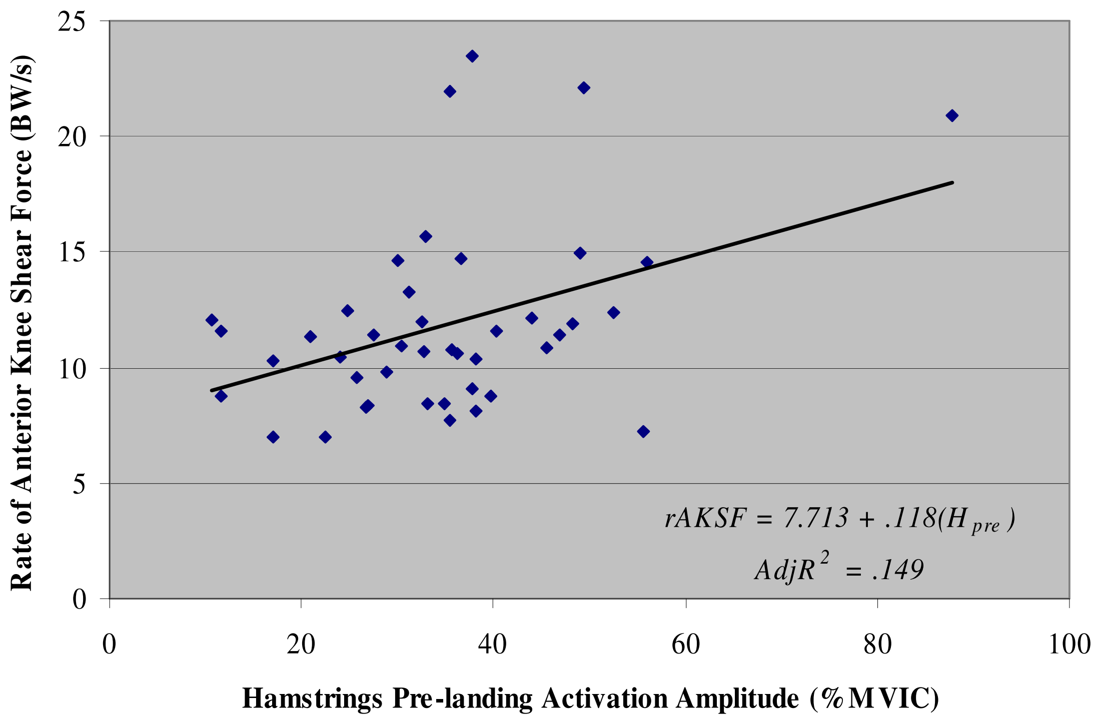
$$rAKSF = 7.713 + .118(H_{pre})$$

Figure 9 graphically depicts this relationship. The full stepwise regression Correlations, Model Summary, ANOVA results, and Coefficient tables can be found in Appendix AA.

<u>Hypothesis III – Prediction of Peak Anterior Knee Shear Force</u>

Examination of the regression coefficients revealed that, H_{post} positively predicted while H_{pre} negatively predicted pAKSF, thereby partially rejecting hypothesis three in regards to H_{post} but supporting hypothesis three in regards to H_{pre}. In the stepwise linear regression, H_{post} entered the model first with the highest zero-order correlation with pAKSF (r = .317), predicting 7.9% of the variance in pAKSF upon landing (AdjR^2, F $change_{(1,41)}$ = 4.583, *P* = .038). Once H_{post} was accounted for, H_{pre} had the highest partial correlation ($r_{partial}$ = -.340) and entered the model next, predicting an additional 8.6% of the variance in pAKSF (AdjR^2, F $change_{(2,40)}$ = 5.231, *P* = .028). Once H_{post} and H_{pre} were accounted for, Q_{pre} had the highest partial correlation ($r_{partial}$ = -.206) and entered the model next, predicting an additional 4.5%. However, the F change for the third predictor was not significant (F $change_{(3,39)}$ = 1.734, *P* = .196). The regression equation for the two predictor model is as follows:

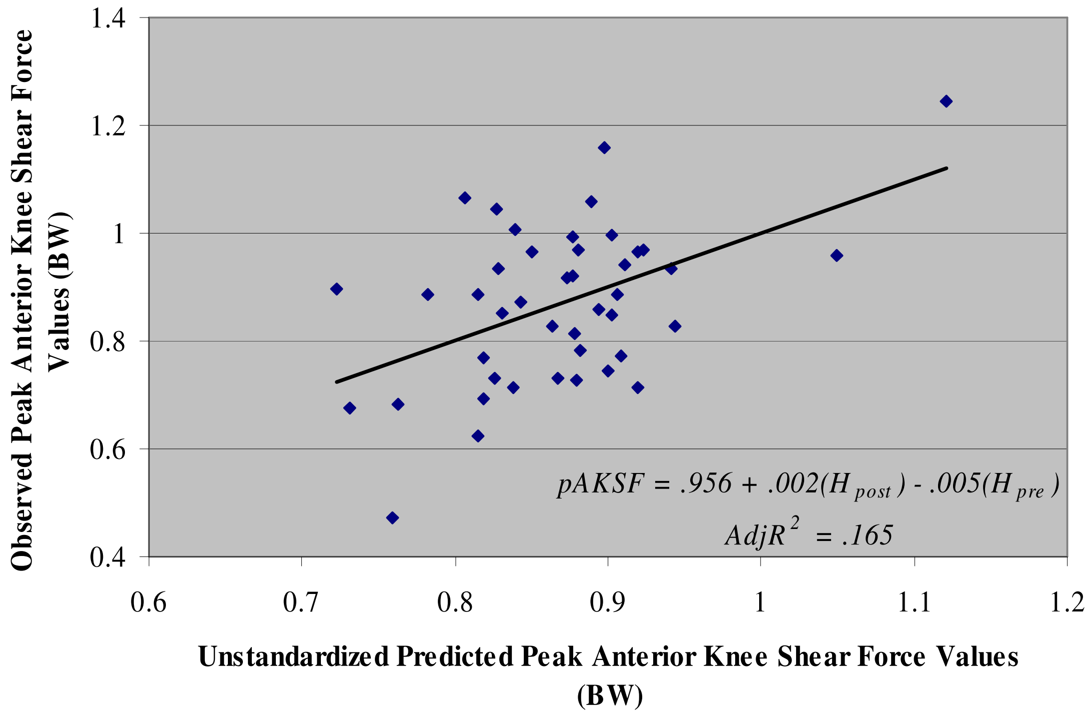
Figure 9. Hamstrings Pre-landing Activation Predicted Rate of Anterior Knee Shear Force



$$pAKSF = .956 + .002(H_{post}) - .005(H_{pre})$$

The graphic relationship between the observed and predicted values of pAKSF can be viewed in Figure 10. The full stepwise regression Correlations, Model Summary, ANOVA results, and Coefficient tables can be found in Appendix BB.

Figure 10. Observed vs. Predicted Peak Anterior Knee Shear Force Values



CHAPTER V

DISCUSSION

This study investigated the collective interactions between hamstrings extensibility, anterior knee laxity, and hamstrings and quadriceps muscle activation strategies as predictors of anterior knee shear forces during single-leg landings in females. While previous studies have examined muscle activation amplitudes (Fagenbaum & Darling, 2003; McNair & Marshall, 1994) and sagittal plane knee joint forces as estimated by inverse dynamics (Hass et al., 2003; Hass et al., 2005) independently in healthy participants during single-leg landings, no published reports were found that have examined these variables in combination while accounting for the modifying effects of hamstrings extensibility and knee laxity. The primary findings were that hamstrings pre-landing activation negatively predicted anterior knee shear force at initial ground contact and positively predicted rate of anterior knee shear force following landing. Hamstrings post-landing activation positively predicted peak anterior knee shear force, and once hamstrings post-landing activation was accounted for, hamstrings pre-landing hamstrings activation negatively predicted peak anterior knee shear force. Anterior knee laxity, hamstrings extensibility, and quadriceps pre- and post-landing activations did not significantly add to the prediction of any of the three dependent variables. This discussion will first focus on a comparison of the descriptive data from this study to those previously reported, followed by a complete assessment of the results obtained for each hypothesis, a discussion of the clinical implications and generalizability of the findings, and the immediate directions for future research.

Quality of the Data Used to Predict Anterior Knee Shear Forces

Before interpreting the stepwise linear regressions, it is first important to examine the data qualitatively to ensure the values obtained reflect the population studied, and that the spread of the data was sufficient to observe the relationships of interest (Pedhazur, 1997). The following section will compare the values obtained in this study with those previously reported in healthy, recreationally active females. For reference, the means and standard deviations for the predictor and dependent variables can be found in Tables 3 and 4, respectively, and histograms depicting the spread of the data for the predictor and dependent variables can be found in Appendices Q through V and W through Y, respectively.

Hamstrings Extensibility

A normal distribution of hamstrings extensibility values were obtained in this study. The methods utilized to measure hamstrings extensibility were modified slightly from those of Blackburn et al. (2004), in which 15 healthy, recreationally active females were studied. Blackburn et al. (2004) positioned the hip in 90° of flexion with a pad against which the participant could rest the posterior thigh while actively extending the knee (Blackburn, Riemann et al., 2004). In the current study, participants were positioned with the hip in 120° of flexion with a horizontal bar placed against the posterior thigh that rotated when a moment exceeding 5Nm was imposed upon it. The change in hip flexion angle made it somewhat difficult to compare values to previous literature. While full active hamstrings extensibility in this study was 138.6 ± 13.1°, mean angles from studies measuring healthy, recreationally active females in 90° of hip flexion ranged from 166.7 to 172.1 (based on the authors' reported acute angles) (Blackburn, Riemann et al., 2004; Nyland et al., 1999; Swanik et al., 2004). While it is not clear if the hip and knee contribute equally to hamstrings extensibility, if the 30° difference in hip flexion were added to the obtuse angles obtained in previous work the corresponding angles of 142.1° (Blackburn, Riemann et al., 2004), 136.7° (Swanik et al., 2004), and 138° (Nyland et al., 1999) are very similar to the mean value in this study.

Anterior Knee Laxity

The histogram for anterior knee laxity (Figure 15) indicates a reasonable range of values were obtained in the sample population, however the majority of the participants fall very close to the mean. It has been suggested that a distribution of predictor variable values that are tightly grouped around the mean may decrease the proportion of the variance explained when using regression (Pedhazur, 1997). Hence, the fact that 20 of 43 participants had a mean anterior knee laxity between 6.5 and 8.5mm is of some concern. On the other hand, a comparison of the present results (KT = 7.0 ± 1.8) to several published reports of anterior knee laxity measured at 134N of force using the KT1000 or KT2000 indicates these findings are not unusual, with the mean and standard deviation being consistent with those previously reported in young, healthy females (range 4.6 ± 1.9 to 7.9 ± 2.0mm) (Rosene & Fogarty, 1999; Rozzi et al., 1999; Trimble et al., 2002; Uhorchak et al., 2003).

Muscle Activation Amplitudes

The histograms for muscle activation (Figures 16 to 19) indicate a reasonable range of values for each of the variables, with one or two participants demonstrating values that were substantially greater than the mean. Casewise diagnostics did not identify these subjects as outliers that would significantly influence the regression results, therefore these values were considered normal and included in the analyses. Mean values obtained in this study are consistent with previous studies reporting muscle activation amplitudes normalized to MVICs during landings in healthy males and females (Colby et al., 2000; Fagenbaum & Darling, 2003; McNair & Marshall, 1994).

Anterior Knee Shear Forces

In this study anterior knee shear forces were estimated by inverse dynamics as the force of the distal femur relative to the tibia or shank reference frame, with a posterior force being positive. The histograms for initial anterior knee shear force (Figure 20) and peak anterior knee shear force (Figure 22) indicate a reasonable spread of data with fairly normal distributions. While the histogram for rate of anterior knee shear force (Figure 21) shows four values that are well above the mean, the casewise diagnostics suggest these values did not have a significant influence on the regression analyses.

Previous studies examining peak anterior knee shear force as estimated by inverse dynamics during landings in healthy females report a range of values consistent with the current findings (Chappell et al., 2002; Hass et al., 2003; Hass et al., 2005; Simpson & Pettit, 1997). The rate of anterior knee shear force as estimated by inverse dynamics has been reported in only one study and ranged from 5.9 to 40.9BW/s during forward landings from 30, 60, and 90% of each participant's maximum forward jump distance (Simpson & Pettit, 1997). The mean rate of anterior knee shear force observed in this study (11.8BW/s) during drop landings with a small forward component (30% of the participants' height) seems reasonably consistent with these findings. While no published reports were found that documented anterior knee shear force at initial contact as estimated by inverse dynamics during landing, one previous report utilized an optimized inverse dynamics knee model to calculate the knee shear force at initial contact of a sudden deceleration on a single leg (Steele & Brown, 1999). Steele and Brown (1999) reported an anteriorly directed force of the tibia relative to the femur of 0.43 ± 0.68BW compared to the -.13 ± 0.19BW estimated solely by inverse dynamics in this study. The optimization model they used estimated the effects of the patellar tendon force and the posterior slope of the tibia (Nissel, Nemeth, & Ohlsen, 1986), both of which would increase the anterior force of the tibia relative to the femur. Therefore, it is not surprising that their reported value of anterior knee shear force at initial contact was greater than that in the current study.

Summary

The mean values for each of the predictor and dependent variables obtained in this study are closely aligned with those previously reported in the literature for similar populations and tasks. While initial anterior knee shear force as estimated purely by inverse dynamics has not yet been reported, comparing the values obtained in this study to those reported in a study that used an optimized inverse dynamics solution indicate these values are reasonable. In general, the data appear to be normally distributed with a few exceptions, mostly confined to the neuromuscular data. While these few values far exceeded the mean value, they did not meet the criteria for exclusion, and were therefore included in all analyses. The one variable of concern is anterior knee laxity. Although laxity values in this sample were normally distributed and consistent with previous literature, the lack of spread in the data may have some influence on its predictive ability. Therefore, with the possible exception of anterior knee laxity, these descriptive findings suggest reasonable values were obtained from the sample population with sufficient spread in the data to draw meaningful conclusions.

Prediction of Anterior Knee Shear Forces

As previously discussed, sagittal plane knee stabilization is provided through both active and passive means. The thigh musculature is thought to play a critical role in controlling anterior knee shear force, with quadriceps activation increasing anterior tibial translation, and hamstrings activation stabilizing the tibia to reduce anterior tibial translation (Hirokawa et al., 1991; Li et al., 1999). Together, quadriceps and hamstring co-activity act to stiffen the joint, leading to increased shear stiffness and reduced anterior tibial translation (Wojtys et al., 2002). However, based on previous work showing reduced muscle stiffness (Blackburn, Padua et al., 2004) and increased hamstrings activation (Shultz, Carcia et al., 2004) in subjects with increased hamstring extensibility and knee laxity, respectively, it was hypothesized that the relative effectiveness of quadriceps and hamstrings muscles in resisting shear loading may be modified due to these anatomical characteristics. Specifically, it was theorized that females with high hamstrings extensibility and/or anterior knee laxity may utilize higher levels of hamstrings activation to control tibial motion and anterior knee shear forces during landings compared to females with low hamstrings extensibility and/or anterior knee laxity. The following discussion will focus on the extent to which the interactive relationships between these anatomical characteristics and thigh muscle activations predicted anterior knee shear forces during single-leg landings in healthy, recreationally active females.

Initial Anterior Shear Force

As estimated in the current study, initial anterior knee shear force represents the sum of the forces imposed on the knee joint when the foot contacts the ground. It is reasonable to expect that the mean value is close to zero (-.13BW), and a slightly negative mean (i.e. a slight posterior shear force) suggests that the body is preparing to experience the anterior knee joint loading expected upon landing (Chappell et al., 2002; Hass et al., 2003; Hass et al., 2005; Simpson & Pettit, 1997). Increased hamstrings activation and decreased quadriceps activation may allow for a posteriorly directed stabilizing force on the tibia, yielding a lower initial anterior knee shear force. Given that that this force may be preparatory in nature, females with greater hamstrings extensibility and/or anterior knee laxity may benefit from greater levels of hamstrings activation and lower levels of quadriceps activation in an effort to reduce initial anterior knee shear force in preparation for the higher anterior knee loads that will occur following landing.

The hypothesis that hamstrings extensibility, anterior knee laxity, and hamstrings pre-landing activation would negatively predict, while quadriceps pre-landing activation would positively predict initial anterior knee shear force was only partially supported. Results indicate that hamstrings pre-landing activation negatively predicted initial anterior knee shear force, however quadriceps pre-activity, knee laxity, nor hamstrings extensibility had any bearing on these findings. While previous studies have reported hamstrings and quadriceps pre-activation amplitudes during landings (Fagenbaum & Darling, 2003; McNair & Marshall, 1994) and have indicated a clear relationship between hamstrings activation and a reduction in anterior tibia translation (Hirokawa et al., 1991; Li et al., 1999) and ACL strain (Fleming, Renstrom, Ohlen et al., 2001), this appears to be the first investigation that has reported a direct relationship between increased hamstrings pre-activation and a reduction in anterior knee shear force as estimated with inverse dynamics during landing.

Given that the knee flexion angle at the time of initial contact ranged from five to 29° in this study, it is likely that increased hamstrings pre-activation levels reduced initial anterior knee shear force primarily through increased knee joint compression (Wojtys et al., 2002), rather than a posterior directed load on the tibia (Herzog & Read, 1993). Evidence suggests that while hamstrings activation may resist anterior tibial translation, these findings have been limited to knee flexion angles greater than or equal to 30° (Hirokawa et al., 1991; Li et al., 1999). At lower knee flexion angles however, increased hamstrings activation will produce a primarily vertical force vector (Herzog & Read, 1993) resulting in increased knee joint compression and stiffness, thereby resisting the translation and shearing of the joint.

Quadriceps activation increases anterior tibial translation in small flexion angles (Hirokawa et al., 1991; Li et al., 1999), and it was therefore expected that lesser quadriceps pre-landing activation would have complimented greater hamstrings pre-landing activation in predicting lesser initial anterior knee shear force. The results do not support this contention, but instead indicate that only hamstrings activation prior to landing contributed to controlling anterior shear force upon ground contact. These findings may, in part, be explained by the nature of the task, and the functional importance of the quadriceps and hamstrings in executing a landing maneuver. Since landing is primarily quadriceps dominated, the quadriceps must be activated at a level sufficient to avoid collapse of the lower extremity (Tillman et al., 2004). The hamstrings are not activated to control this collapse, but are mainly activated to control hip flexion (Tillman et al., 2004), and, as the present findings suggest, to control the sagittal forces at the knee. While hamstrings and quadriceps pre-landing activation were moderately correlated in this study (r = .443, $P = .003$), the lack of a stronger relationship suggests that the possibility of quadriceps and hamstrings acting independently prior to landing does exist. Hence, regardless of the quadriceps activation prior to landing, only when the hamstrings activate at a high level will the sagittal forces imposed upon the knee be controlled at ground contact.

From the current data, it is difficult to determine why anatomical characteristics did not contribute to the prediction of initial anterior knee shear force. The results of the regressions and the lack of correlations between initial anterior knee shear force and hamstrings extensibility (r = .037, $P = .815$) or anterior knee laxity (r = .189, $P = .225$), illustrate that little to no relationships were identified between the anatomical characteristics and initial anterior knee shear force. While individual correlations may not reflect the interaction of the predictor variables as they contribute to the control of initial anterior knee shear force, the lack of correlations between muscle activation and hamstrings extensibility or anterior knee laxity further illustrate the lack of dependence of muscle activation strategies on these anatomical characteristics.

It is possible that the influence of hamstrings extensibility on active muscle stiffness (Blackburn, Padua et al., 2004) and neuromuscular control of joint stability may be minimized during a task such as landing where the hamstrings are already activated to a high level well before ground contact (Fagenbaum & Darling, 2003; McNair & Marshall, 1994). It is difficult to support this contention as published reports to date have yet to examine the relationship between hamstrings extensibility and hamstrings activation strategies during functional tasks such as landing. Similarly, while females with greater anterior knee laxity may prepare for a single-leg weight-bearing perturbation by increasing hamstrings pre-activation (Shultz, Carcia et al., 2004), this relationship may not be realized during a landing maneuver, where the need to pre-activate to control joint forces during landing (Cowling & Steele, 2001) may exceed that needed to control excess tibio-femoral motion during single leg stance. However, the grand mean for hamstrings pre-perturbation amplitudes across subjects [high (>7mm) and low (<5mm) anterior knee laxity] (~30%) (Shultz, Carcia et al., 2004) is only slightly lower than the hamstrings pre-activation amplitudes in this study (35%) and other landing studies [41% (McNair & Marshall, 1994) and ~45% (Fagenbaum & Darling, 2003)]. Hence, while the theories suggested for these results are plausible, the present findings and published reports to date do not readily explain why anatomical characteristics and thigh muscle activation did not interact to control initial anterior knee shear force in this study.

Rate of Anterior Knee Shear Force

As estimated in this study, rate of anterior knee shear force represents the relative change in anterior knee shear force (peak minus initial), divided by the amount of time it takes to achieve peak anterior knee shear force (tpAKSF):

$$rAKSF = (pAKSF - iAKSF) / tpAKSF$$

This calculation dictates that a higher rate of anterior knee shear force may result from some combination of a low initial force, shorter time to peak force, and/or a high peak force. Hence, the rate of anterior knee shear force in this study represents either the participant's ability to control the magnitude of the change in anterior knee shear force and/or the time course to reach peak anterior knee shear force following landing.

Based on previous work it was hypothesized that anterior knee laxity would negatively predict the rate of anterior knee shear force secondary to an ability of females with high knee laxity to increase hamstrings activation to control the magnitude of change in force by reducing the peak force (Shultz, Carcia et al., 2004). Conversely, it was hypothesized that hamstrings extensibility would positively predict the rate of anterior knee shear force due to an inability of females with high hamstrings extensibility to produce muscle stiffness to control the magnitude of change in anterior knee shear force (Blackburn, Padua et al., 2004). The hypothesis that hamstrings extensibility and quadriceps activation would positively predict, while knee laxity and hamstrings activation would negatively predict the rate of anterior shear force was rejected, with findings indicating that hamstrings pre-landing activation was the sole predictor of rate of anterior knee shear force. However, the nature of this relationship was such that hamstrings pre-landing activation positively (rather than negatively) predicted the rate of anterior knee shear force.

Given the nature of the calculation used to estimate rate of anterior knee shear force, and the established relationship between hamstrings activation and control of anterior movement of the tibia (Hirokawa et al., 1991; Li et al., 1999), these findings suggest that greater hamstrings activation may have increased the joint stiffness upon landing and resulting in greater change in force and a shorter time to peak force. The findings from hypothesis one indicate that hamstrings pre-landing activation predicted a reduction in the initial anterior knee shear force. This result may have contributed to a greater change in force, thereby increasing the rate of anterior knee shear force. Furthermore, hamstrings pre-landing activation was somewhat negatively related to the time to peak anterior knee shear force ($r = -.23$, $P = .14$) (Appendix CC), contributing further to the positive predictive relationship between hamstrings pre-landing activation and the rate of anterior knee shear force. Therefore, hamstrings pre-landing activation was a positive predictor of the rate of anterior knee shear force via its combined relationships with an increase in the change in anterior force and a reduction in the time to peak anterior force. While it is not readily apparent if this increased rate of force is clinically beneficial to controlling the knee joint in the sagittal plane, it is unlikely that it would result in poor sagittal plane knee control unless it were accompanied by a greater peak anterior knee shear force.

Given the quadriceps role in producing increased anterior tibial translation (Hirokawa et al., 1991; Li et al., 1999), it is somewhat surprising that quadriceps pre-landing activation did not contribute to the prediction of rate of anterior knee shear force by altering the length of time to peak force. As was the case with initial anterior knee shear force, the lack of predictive ability of the quadriceps to determine rate of anterior knee shear force may be due to the landing task being quadriceps dominated, leaving hamstring activation to ultimately determine joint stability. Hence, these data further support the role of hamstrings pre-activation amplitude as a primary factor in controlling anterior knee shear force during a landing task.

It is easier to explain the lack of the role of post-landing hamstrings and quadriceps activations in determining the rate of anterior knee shear force. With the rate of anterior knee shear force being largely dependent on the time to peak anterior knee shear force, it may be that post-landing activation occurred too late to affect the rate of anterior knee shear force. An electromechanical delay exists between when muscle activation occurs and when muscle force generation is sufficient to influence joint dynamics (Winter & Brookes, 1991). Electromechanical delay in females has been reported to be approximately 45ms, however it is important to acknowledge that these findings are limited to relaxed muscle conditions (Winter & Brookes, 1991). While the author was not able to locate data indicating the magnitude of electromechanical delay of a previously contracted muscle during a closed-chain activity, an electromechanical delay of 40ms has previously been used to time shift sEMG data during landing when attempting to observe relationships between muscle activation and net joint moments (McNitt-Gray, Hester, Mathiyakom, & Munkasy, 2001) Considering the time to peak force observed in this study (89 ± 19ms), it is plausible that muscle activations post-landing did not generate additional force early enough to contribute significantly to the rate of anterior knee shear force. More work is needed to determine the most appropriate analyses to fully understand the relationship between muscle activation and joint kinetics.

Given similar levels of muscle activation, it was expected that hamstrings extensibility would positively predict and anterior knee laxity would negatively predict the rate of anterior knee shear force. As with initial anterior shear force, hamstrings activation appears to work independent of anatomical characteristics to influence rate of anterior knee shear force through control of both the initial force and the time to peak force following landing. The lack of correlations between anatomical characteristics and rate of anterior knee shear force or thigh muscle activation once again suggest that the level of hamstrings activation needed to control the rate of knee joint forces during landing may exceed that which is necessary to control knee joint forces during a single-leg weight-bearing perturbation.

<u>Peak Anterior Knee Shear Force</u>

As estimated in this study, peak anterior knee shear force represents the highest value of the sum of the forces imposed on the knee joint following ground contact (Robertson et al., 2004). Clinically, this measure may represent a participant's ability to control the highest forces imposed on the knee following a landing. A lesser peak anterior knee shear force may be reflective of an ability to control the great forces imposed on the knee joint following landing, while a greater peak shear may indicate the opposite. As with initial shear force, it was expected that quadriceps activation would positively predict and hamstrings activation would negatively predict peak shear force secondary to the effects of thigh muscle activation on the control of tibial translation (Hirokawa et al., 1991; Li et al., 1999). Furthermore, it was expected that hamstrings extensibility would positively predict and anterior knee laxity would negatively predict peak force secondary to their proposed interactions with control of tibial motion and force via altered hamstrings function (Blackburn, Padua et al., 2004; Shultz, Carcia et al., 2004).

As was the case with both initial and rate, peak anterior knee shear force was primarily influenced by hamstrings activation. However, in this instance, hamstrings post-landing activation positively predicted, while hamstrings pre-landing activation negatively predicted peak anterior knee shear force. Previous reports have indicated that peak anterior knee shear forces are greater in females than in males (Chappell et al., 2002) and greater in post-pubescent than pre-pubescent females during forward (Hass et al., 2003) but not vertical and lateral landings (Hass et al., 2005). However, this appears to be the first published report that has described the relationship between hamstrings activation and peak anterior knee shear force following a landing.

While the positive relationship between post-landing hamstrings activation and peak anterior shear force may seem counterintuitive, it is likely that the relationship observed may be responsive in nature. As previously discussed, the natural delay that occurs between muscle activation and force generation (Winter & Brookes, 1991) may be too slow to contribute to the control of anterior shear forces that peak within 89 ± 19ms of ground contact. Therefore, the positive relationship between hamstrings post-landing activation and peak anterior knee shear force is likely the result of the muscle responding to, rather than controlling, the peak force.

The interesting finding in the regression analysis for predicting peak anterior shear force is that hamstrings post-landing activation entered the model first. Hamstrings pre-landing activation alone was not significantly correlated to peak anterior knee shear force ($r = .026$, $P = .870$), and it was only after hamstrings post-landing activation was accounted for that hamstrings pre-landing activation had a significant negative relationship with peak anterior knee shear force ($r_{partial} = -.340$, $P = .028$). While it is counter-intuitive to think of post-landing activation entering the regression model prior to pre-landing activation, this relationship was such that if a female were to increase activation of their hamstrings both prior to and following landing, the peak anterior knee shear force would be less than if the hamstrings were not highly active prior to landing. In other words, females who do not pre-activate the hamstrings with sufficient amplitudes will have hamstrings responses to landing that will be too late to reduce peak anterior knee shear force. As before, the relationship between hamstring activation and the control of peak anterior shear force was not dependent on the level of quadriceps activation or the presence of increased knee laxity and/or hamstring extensibility likely due to the previously suggested theories.

Collective Findings as They Relate to Previous Reports

The collective findings of this investigation indicate that increased hamstrings activation was the primary factor in controlling anterior knee shear forces throughout a single-leg landing in healthy, recreationally active females. Increased hamstrings pre-landing activation predicted a reduction in anterior knee shear force regardless of the participants' hamstrings extensibility, anterior knee laxity, or quadriceps muscle activation strategies. The fact that quadriceps activation had no bearing on the prediction of anterior knee shear forces suggests that the level of hamstrings to quadriceps activation may not be as critical of a factor in controlling anterior knee shear force prior to or following landing as previously thought (Croce et al., 2004). Rather, it is the absolute level of hamstrings pre-activation in preparation for landing that seems to be critical.

While published reports investigating the direct relationship between anterior knee shear forces as estimated by inverse dynamics and hamstrings activation during landings are lacking, investigations have shown similar relationships between hamstrings activation and net knee joint moments as estimated by inverse dynamics (Doorenbosch & Harlaar, 2003; Doorenbosch, Welter, & Van Ingen Schenau, 1997; Prilutsky, Gregor, & Ryan, 1998). While it is recognized that the estimation of moments through inverse dynamics is not the same as that for joint forces, it is likely they are influenced by the same factors. As with force, net joint moments are estimated with the use of the participant anthropometrics (in this case the inertial properties of the segments about the joint), ground reaction force data, and kinematic data (in this case the angular acceleration between the segments of the joint) (Robertson et al., 2004; Winter, 1990).

Further, the individual contributions of muscle, capsular, and ligamentous forces are not calculated, but theoretically may influence the angular acceleration and as such the joint moments.

While conclusions from the current study suggest that hamstrings activation is more important than quadriceps activation in controlling forces at the knee, the analyses in previous studies (Doorenbosch & Harlaar, 2003; Doorenbosch et al., 1997; Prilutsky et al., 1998) do not allow the reader to determine if the hamstrings or quadriceps were the primary factor in controlling the net moments. In one study, the authors report a strong relationship ($r = .986 \pm .004$) between the difference in rectus femoris and hamstrings activation (rectus – hamstrings) and the difference in the extension moment at the knee and hip joints (knee - hip) (Doorenbosch et al., 1997). Studies since have illustrated that the activation ratio of the hamstrings and quadriceps may be used to predict the net knee joint moment during a vertical jump (Doorenbosch & Harlaar, 2003), and that increased hamstrings activation is related to increased knee flexion moments during walking and running (Prilutsky et al., 1998). None of these investigations analyzed the influence of hamstrings and quadriceps activation as multiple independent predictors of the net joint moments. Hence, it can not be concluded which of these muscles was primarily responsible for the change in net moments as was illustrated with the hamstrings and anterior knee shear forces in this study.

Clinical Implications of the Findings

The current findings give insight as to how females utilize their hamstrings to influence sagittal forces during functional tasks such as landings. While ACL injury risk was not measured directly in this study, these findings lend support to the fact that females who are able to pre-activate their hamstrings selectively prior to ground contact may improve sagittal knee stabilization during high velocity athletic tasks such as single-leg landings. Therefore, ACL injury prevention programs and ACL-deficient or ACL-reconstruction rehabilitation programs may benefit from a focus on enhancing hamstrings pre-activation during functional tasks. This contention is supported by the use of exercises that emphasize hamstrings strengthening (Hewett, Lindefeld, Riccobene, & Noyes, 1999; Hewett et al., 1996) and/or proprioceptive training of the hamstrings (Caraffa, Cerulli, Projetti, Aisa, & Rizzo, 1996; Myklebust et al., 2003) as components of ACL injury prevention programs that have been shown to reduce injury risk. This is also supported by post ACL-injury rehabilitation programs that focus on selective hamstrings activation during perturbation training programs in athletes who are ACL-deficient (Chmielewski, Rudolph, & Snyder-Mackler, 2002; Fitzgerald, Axe, & Snyder-Mackler, 2000), and those that focus on both hamstrings strength and proprioception training in patients following ACL-reconstruction (Nyland, Currier, Ray, & Duby, 1993; Shelbourne & Davis, 1999).

Generalizability of the Findings

In order to apply the findings of this study to future studies and to clinical practice, it is first important to assess the ability to generalize these findings. The single-leg landings performed in this study were quite novel, mostly confined to one plane of motion, and highly controlled to allow the assessment of the proposed relationships. It is important to note that on the field of play, where ACL injuries occur, athletes move in multiple planes of motion, and factors such as goal orientation and unexpected perturbations from other athletes or objects likely mediate the relationships observed, possibly to the extent that anterior knee laxity and hamstrings extensibility may play a role.

It should be also acknowledged that regressions can only provide information about the relationships between the variables included in the analyses, which may differ considerably once other factors are included or excluded. For example, additional anatomical characteristics may play a role in influencing the proposed relationships during this and other athletic tasks, and/or females that participate in other levels or types of activities may exhibit different relationships than those observed. It is also important to emphasize that sEMG activation is not directly reflective of muscle force or of the muscle's ability to influence the task, and that forces estimated through inverse dynamics are only reflective of the minimum sum of forces experienced by the knee joint (Winter, 1990). Further, these anterior knee shear forces do not represent the amount of force being imposed on the ACL during these landings, and therefore this study did not directly assess the risk of ACL injury, but rather the relative contributions to anterior knee shear force during single-leg landings in females. Finally, only females were studied. While this delimitation was important to control for other sex confounding factors, it is unknown whether similar relationships would have been observed between the variables of interest in a male population.

Immediate Directions for Future Research

The immediate directions for future research should include continued investigations that identify variables that predict or explain parameters of knee stabilization in at risk populations. The current research may benefit from a more comprehensive assessment of anatomical characteristics that affect the sagittal plane, such as genu recurvatum and pelvic inclination. In addition to shear forces, future investigations may wish to explore pre- and post landing muscle activation latencies, knee flexion angles, and joint moments as outcome variables, which may further clarify their role in sagittal plane joint stabilization. If neuromuscular activation strategies are to be considered, evidence suggests a time shift in the sEMG data to reflect electromechanical delay may more accurately assess the temporal relationships between anatomical characteristics, neuromuscular activation strategies and joint kinetics (McNitt-Gray et al., 2001). Finally, it is unknown if the neuromuscular strategies in this study, and their relationship to knee forces, would hold for other tasks that involve frontal and transverse plane movements such as cutting and pivoting. Continued development of relevant prediction models that elucidate the understanding of knee joint stabilization in multiple planes during a variety of dynamic functional tasks is needed. Only upon refinement of these prediction models over time, will clinicians have the knowledge necessary to effectively intervene to reduce injury risk and improve rehabilitation outcomes.

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APPENDIX A. Institutional Review Board Consent Form

THE UNIVERSITY OF NORTH CAROLINA
GREENSBORO

CONSENT TO ACT AS A HUMAN PARTICIPANT: LONG FORM

Project Title: The contributions of sagittal knee laxity, joint angle, and muscle activation to anterior knee shear force during single-leg landings in females.

Project Director: Thomas C. Windley

Participant's Name: ____

DESCRIPTION AND EXPLANATION OF PROCEDURES:
The purpose of the study is to establish the contributions of leg structure, knee joint motion, and muscle activation to knee forces during a single-leg landing. In order to qualify for this investigation, you must be recreationally active (participate in physical activity for a minimum of 90 minutes per week) and have no injuries to the legs or feet in the past six months, no injuries to the legs or feet that have resulted in ligament tears, and no surgeries on the legs or feet. Additionally, if you are, or suspect that you may be pregnant, you will not be allowed to participate in this study, due to hormonal changes that may confound our knee laxity results. Testing will consist of one session that will last approximately two hours.

Height, weight and measures of leg structure will be measured and recorded. An age and physical activity history will also be taken. Once these preliminary measures are taken the landing task will be instructed and demonstrated to you. You will then be allowed to practice the landings until you are comfortable with the task. Then four pairs of surface electrodes will be secured on the skin over the muscles on the front and back of your thigh and on your calf. You will then be placed in a seat with your knee fixed at a 30 degree angle. You will then perform three maximal pushes and pulls of your leg against a fixed pad for five seconds each. Then you will perform three maximal pushes of your foot down against a fixed resistance for five seconds. Next, four sensors will be secured to you (3 on one leg, and one on the back of your pelvis) using double-sided tape and /or Velcro straps and athletic tape. Formal data collection of the landings will then begin.

Landing Measures: You will drop off a 30 cm box onto a force plate until we obtain 5 good trials. Each time you will be instructed to hold your hands on your hips throughout the whole trial, drop, not jump, onto the force plate on your dominant leg, and hold your balance on that leg for one second. The previously attached EMG electrodes will record muscle activity while the sensors will record joint movement, and the force plate will record the forces placed on your body during the landing.

The entire data collection process will take approximately two hours.

RISKS AND DISCOMFORTS:
There are no anticipated risks. There is a small possibility that you could have a mild muscle strain during the maximum testing, however you are positioned to prevent that from occurring and in previous like research no one has sustained this injury. There is a possibility you may lose your balance during the testing and subsequently suffer a strain, sprain or contusion. However, to minimize this risk, there will be a safety handrail and an investigator nearby to provide protection should you begin to lose your balance. If at any time the testing causes you any discomfort or concern, please notify the investigator immediately.

POTENTIAL BENEFITS:
There are no direct benefits to you for participating in the study. The study may help us better understand the risk factors associated with knee ligament injuries.

CONSENT:
By signing this consent form, you agree that you understand the procedures and any risks and benefits involved in this research. You are free to refuse to participate or to withdraw your consent to participate in this research at any time without penalty or prejudice; your participation is entirely voluntary. Your privacy will be protected because you will not be identified by name as a participant in this project.

The research and this consent form have been approved by the University of North Carolina at Greensboro Institutional Review Board, which insures that research involving people follows federal regulations. Questions regarding your rights as a participant in this project can be answered by calling Mr. Eric Allen at (336) 256-1482. Questions regarding the research itself will be answered by Thomas C. Windley by calling (336) 334-3039 or Dr. Sandra J. Shultz by calling (336) 334-3027. Any new information that develops during the project will be provided to you if the information might affect your willingness to continue participation in the project.

By signing this form, you are agreeing to participate in the project described to you by Thomas C. Windley. A copy of this consent form will be provided to you.

___________________________________ ______________

Participant's Signature* Date

APPENDIX B. Activity Questionnaire

Activity and Injury Questionnaire: The contributions of sagittal knee laxity, joint angle, and muscle activation to anterior knee shear force during single-leg landings in females

Subject ID#:_______________ Date:________________

Age:_________ Height:__________ Weight______________

1. Have you had an injury to either leg in the past 6 months that has limited your normal activities for more than one day? Yes No

2. Have you ever ruptured or torn completely and ligaments in your feet, ankles, knees, or hips? Yes No

3. Have you ever had surgery on either one of your legs?
Yes No

4. Do you have any medical conditions that prohibit you from exercise participation?
Yes No

4. If not, are you currently participating in regular exercise?
Yes No

5. If yes, how many hours do you exercise per week?__________

6. How many of these hours are devoted to jumping and landing activities?___________

7. What types of jumping and landing activities have you been regularly engaging in over the past six months (ie. Basketball, volleyball, etc.)?
____

8. Prior to the last six months did you have any experience in jumping and landing activities? Yes No

9. If yes, what activities were they?____

10. How many years have you engaged in those activities? _________________________
____

11. Over those years, approximately how many months/year? ______________________,

days/week?________________________, hours/session?____________________________

12. What was the date that you began your last menstrual cycle?_____________________

APPENDIX C. Data Collection Sheet

Demographic Measures

Subject ID: ____________________

Sub ID	Age	Wt (kg)	Ht (cm)	Drop Leg

Date: ____________________

30% of Ht: ____________

Anatomical Measures Right

Standing

Par	StQr	TFAr	FLr	TLr	NDr

Prone Supine

Har	KTr	GRr	TTr	HSr

Anatomical Measures Left

Standing

Pal	StQl	TFAl	FLl	TLl	NDl

Prone Supine

Hal	KTl	GRl	TTl	HSl

APPENDIX D_1. Raw Demographic Variable Data

Subject	Age (yrs)	Weight (kg)	Height (cm)	Preferred Leg	Hours Exercise/Week	Hours Landing Activity/Week
1	19	60.6	163.9	Right	2.0	2.0
2	22	87.9	170.6	Right	1.5	0.5
3	30	42.8	153.6	Right	2.5	0.8
4	26	51.7	155	Right	2.3	0.0
5	22	59.6	165.9	Left	3.5	3.5
6	20	82.7	168.4	Left	10.0	0.0
7	22	48.8	155	Left	12.0	0.0
8	19	79	169.2	Left	3.0	0.0
9	20	54.4	158.4	Right	4.0	2.5
10	20	63.1	173.9	Right	12.0	6.0
11	30	69.8	172.9	Right	2.0	0.0
12	20	51.2	169.2	Right	4.0	0.0
13	24	76.5	175.5	Left	8.0	0.0
14	20	54.6	162.5	Right	7.0	2.0
15	19	66.7	173.4	Right	11.0	4.0
16	19	65.4	157.2	Right	5.0	3.0
17	21	67.5	176.8	Right	8.0	1.5
18	24	54.2	161.5	Right	5.0	0.0
19	21	45.8	150	Right	7.0	0.0
20	20	85.2	167.6	Right	2.0	0.0
21	19	57.8	160.6	Right	2.5	0.0
22	30	65.8	171.7	Right	4.0	0.0
23	21	74	169.1	Right	4.5	2.0
24	22	65.6	161	Left	12.5	0.0
25	25	62.6	171.8	Right	10.0	0.5

Subject	Age (yrs)	Weight (kg)	Height (cm)	Preferred Leg	Hours Exercise/Week	Hours Landing Activity/Week
26	21	47.7	161.9	Right	3.0	1.0
27	20	52.2	156.8	Right	13.0	3.5
28	20	34.9	160.8	Left	5.0	0.0
29	26	51.4	157.4	Right	2.0	0.0
30	19	54.3	161	Right	4.5	2.0
31	19	54.8	167.9	Left	3.0	0.0
32	25	61.5	166.1	Right	5.0	0.0
33	21	101.3	159.8	Right	8.0	3.0
34	21	86.2	170.4	Right	4.5	0.0
35	19	67.6	167.6	Left	3.5	0.0
36	21	75.7	169.8	Right	6.0	0.0
37	23	51.5	170.8	Right	1.5	0.5
38	22	56.5	156.3	Left	6.0	2.5
39	21	62.8	163.8	Right	3.0	0.0
40	23	75.9	178.3	Right	3.5	0.5
41	20	61.9	167.1	Right	3.5	0.0
42	19	53	162	Left	3.5	1.0
43	20	53.2	155.8	Right	4.0	1.0
44	20	58.2	156.8	Right	6.0	1.5
45	19	66	166.7	Right	7.5	3.0

APPENDIX D_2. Raw Predictor Variable Data

Subject	KT (mm)	HS (deg)	H_{pre} (%MVIC)	H_{post} (%MVIC)	Q_{pre} (%MVIC)	Q_{post} (%MVIC)
1	4.7	137.3	0.44	0.33	0.36	1.24
2	5.2	128.0	2.43	5.56	1.12	2.61
3	7.0	144.0	0.17	0.11	0.46	0.52
4	8.0	123.0	0.47	0.45	0.66	1.22
5	9.3	140.3	0.28	0.21	0.57	2.42
6	7.2	149.3	0.56	0.67	0.41	1.19
7	7.8	147.0	0.33	0.40	0.23	0.42
8	9.3	132.7	0.36	0.52	0.46	1.24
9	12.0	131.3	0.40	0.33	0.44	1.50
10	7.3	155.0	0.38	0.21	0.30	2.02
11	4.3	136.0	0.56	0.46	0.50	0.53
12	10.0	119.3	0.38	0.57	0.47	1.23
13	8.0	106.0	0.40	0.27	0.76	1.27
14	6.2	125.0	0.36	0.24	0.33	1.02
15	5.7	160.3	0.35	0.52	0.33	0.61
16	4.0	153.0	0.30	0.18	0.29	0.56
17	5.7	132.0	0.29	0.42	0.31	0.62
18	7.8	141.3	0.38	0.41	0.59	2.25
19	8.0	126.0	0.21	0.15	0.30	0.47
20	8.0	155.7	0.17	0.25	0.38	2.21
21	6.0	133.0	0.38	0.65	0.35	0.84
22	3.8	118.3	0.48	0.30	0.83	1.19
23	7.5	141.0	0.27	0.25	1.57	1.97
24	5.0	146.7	0.31	0.33	0.33	0.46
25	4.5	140.7	0.12	0.17	0.48	0.61

Subject	KT (mm)	HS (deg)	H_{pre} (%MVIC)	H_{post} (%MVIC)	Q_{pre} (%MVIC)	Q_{post} (%MVIC)
26	6.8	123.0	0.11	0.15	0.30	0.69
27	7.2	137.7	0.23	0.33	0.32	0.33
28	5.7	141.7	0.29	0.12	0.45	2.82
29	8.7	158.3	0.33	0.24	0.39	0.96
30	7.7	123.7	0.36	1.29	0.53	0.91
31	10.0	134.7	0.37	0.34	0.52	1.17
32	7.5	151.0	0.33	0.33	0.52	1.11
33	7.0	162.7	0.52	1.54	1.18	2.44
34	4.0	149.3	0.31	0.32	0.41	0.64
35	7.3	143.3	0.24	0.17	0.30	0.64
36	7.5	151.3	0.26	0.20	0.38	0.78
37	8.7	149.3	0.33	0.20	0.50	0.73
38	6.2	151.3	0.25	0.18	0.32	0.70
39	7.8	130.3	0.46	0.89	0.56	1.73
40	8.2	139.0	0.35	0.61	0.50	0.76
41	4.0	138.7	0.49	0.58	0.39	0.80
42	6.0	141.0	0.49	0.92	0.46	0.75
43	6.8	140.0	0.27	0.31	0.42	0.70
44	8.7	126.7	0.88	3.22	1.42	5.24
45	5.0	113.3	0.12	0.20	0.65	1.33

APPENDIX D_3. Raw Dependent Variable Data

Subject	iAKSF (BW)	tpAKSF (ms)	rAKSF (BW/s)	pAKSF (BW)
1	-0.23	106.6	12.11	1.06
2	0.23	62.4	14.08	1.11
3	0.34	90.8	6.96	0.97
4	0.12	66.6	11.42	0.88
5	-0.28	87.2	11.43	0.72
6	-0.19	121.4	7.25	0.69
7	-0.13	92.8	10.68	0.86
8	-0.03	78.8	10.75	0.81
9	-0.06	94	8.78	0.77
10	-0.15	95.8	8.11	0.62
11	-0.57	71.4	14.57	0.47
12	-0.04	84.4	9.11	0.73
13	-0.23	78.6	11.58	0.68
14	-0.07	94.4	10.64	0.93
15	-0.05	94.2	8.45	0.74
16	-0.32	88	14.63	0.97
17	-0.23	102.2	9.79	0.77
18	-0.08	77.8	10.40	0.73
19	-0.12	104	11.33	1.06
20	0.12	82.6	10.31	0.97
21	-0.22	49.6	23.45	0.94
22	-0.20	92	11.91	0.90
23	-0.25	112.4	8.25	0.67
24	-0.23	104.8	10.96	0.92
25	-0.02	112.8	8.76	0.97

Subject	iAKSF (BW)	tpAKSF (ms)	rAKSF (BW/s)	pAKSF (BW)
26	-0.10	86.2	12.03	0.93
27	0.16	79.8	6.97	0.71
28	-0.85	75.6	36.98	1.95
29	-0.04	87.8	11.97	1.01
30	0.05	117.8	7.72	0.96
31	-0.24	87.4	14.73	1.05
32	0.10	49.2	15.67	0.87
33	-0.44	102.2	12.36	0.83
34	0.11	54.4	13.29	0.83
35	-0.26	119.8	10.49	0.99
36	-0.15	91.6	9.56	0.73
37	-0.05	110.8	8.44	0.89
38	-0.22	91.2	12.49	0.92
39	0.29	51.4	10.89	0.85
40	-0.38	69.8	21.98	1.16
41	-0.21	48	22.10	0.85
42	-0.26	84.2	14.93	1.00
43	-0.19	116.8	8.34	0.78
44	-0.52	84.4	20.86	1.24
45	-0.31	103.8	11.54	0.89

APPENDIX E. Casewise Diagnostics

Participants 2 and 28 Removed Due to Violation of Outlier Criteria

	Dependent Var	Leverage	Z-residual	Cook's D
Hypothesis 1	iAKSF			
	Participant 2	0.83*	0.89	7.92**
	Participant 28	0.01	-3.09*	0.09
Hypothesis 2	rAKSF			
	Participant 2	0.13	-0.39	0.01
	Participant 28	0.21	3.45*	1.15**
Hypothesis 3	pAKSF			
	Participant 2	0.13	0.61	0.02
	Participant 28	0.21	3.72*	1.34**

*Subject flagged for possible influence on the regression

**High Cook's D Value Justified Removal of Flagged Cases

Casewise Diagnostics for Prediction of iAKSF

Subject	Leverage*	Z- residual**	Cook's D
1	0.04	-0.31	0.00
2	0.83*	0.89	7.92
3	0.01	2.33**	0.05
4	0.01	1.19	0.01
5	0.04	-0.79	0.01
6	0.02	-0.49	0.00
7	0.03	-0.33	0.00
8	0.04	0.17	0.00
9	0.19	-0.32	0.01
10	0.02	-0.32	0.00
11	0.05	-1.79	0.07
12	0.07	0.05	0.00
13	0.02	-0.28	0.00
14	0.01	0.25	0.00
15	0.02	0.41	0.00
16	0.07	-0.62	0.01
17	0.02	-0.37	0.00
18	0.01	0.27	0.00
19	0.02	-0.11	0.00
20	0.02	1.11	0.01
21	0.01	-0.41	0.00
22	0.11	0.39	0.01
23	0.43*	0.66	0.16
24	0.03	-0.28	0.00
25	0.07	1.06	0.03

*Leverages >3(4 + 1)/45 = .33 were flagged

**Z-residuals >|2| were flagged

Subject	Leverage*	Z- residual**	Cook's D
26	0.02	0.21	0.00
27	0.01	1.28	0.02
28	0.01	-3.09**	0.09
29	0.03	0.16	0.00
30	0.00	0.84	0.00
31	0.06	-0.82	0.02
32	0.00	1.11	0.01
33	0.13	-0.77	0.03
34	0.06	1.48	0.06
35	0.01	-0.70	0.00
36	0.01	-0.12	0.00
37	0.02	0.24	0.00
38	0.02	-0.36	0.00
39	0.01	1.87	0.03
40	0.01	-1.23	0.01
41	0.06	-0.16	0.00
42	0.01	-0.57	0.00
43	0.00	-0.21	0.00
44	0.23	-1.39	0.22
45	0.07	-0.16	0.00

*Leverages >3(4 + 1)/45 = .33 were flagged

**Z-residuals >|2| were flagged

Casewise Diagnostics for Prediction of rAKSF

Subject	Leverage*	Z- residual**	Cook's D
1	0.06	-0.77	0.01
2	0.13	-0.39	0.01
3	0.02	-0.62	0.00
4	0.02	0.21	0.00
5	0.08	-0.59	0.01
6	0.00	-1.12	0.01
7	0.04	0.11	0.00
8	0.04	0.09	0.00
9	0.18	0.02	0.00
10	0.10	-1.72	0.11
11	0.06	0.43	0.00
12	0.07	-0.09	0.00
13	0.04	0.33	0.00
14	0.02	-0.61	0.00
15	0.02	-0.84	0.01
16	0.07	0.07	0.00
17	0.02	-0.60	0.00
18	0.04	-0.96	0.02
19	0.03	0.33	0.00
20	0.09	-1.18	0.05
21	0.01	2.19**	0.05
22	0.11	-0.34	0.01
23	0.43	0.00	0.00
24	0.04	-0.33	0.00
25	0.05	-0.84	0.01

*Leverages >3(6 + 1)/45 = .47 were flagged
**Z-residuals >|2| were flagged

Subject	Leverage*	Z- residual**	Cook's D
26	0.01	0.05	0.00
27	0.03	-0.62	0.01
28	0.21	3.45**	1.15
29	0.03	0.33	0.00
30	0.01	-0.56	0.00
31	0.07	1.18	0.04
32	0.00	0.89	0.01
33	0.12	-0.14	0.00
34	0.06	-0.11	0.00
35	0.02	-0.12	0.00
36	0.01	-0.30	0.00
37	0.04	-0.09	0.00
38	0.01	0.02	0.00
39	0.01	-0.48	0.00
40	0.03	2.59**	0.09
41	0.06	1.56	0.06
42	0.01	0.63	0.00
43	0.01	-0.57	0.00
44	0.46	0.03	0.00
45	0.03	-0.52	0.00

*Leverages >3(6 + 1)/45 = .47 were flagged

**Z-residuals >|2| were flagged

Casewise Diagnostics for Prediction of pAKSF

Subject	Leverage*	Z- residual**	Cook's D
1	0.06	0.30	0.00
2	0.13	0.61	0.02
3	0.02	0.82	0.01
4	0.02	0.29	0.00
5	0.08	-1.43	0.07
6	0.00	-1.10	0.01
7	0.04	0.13	0.00
8	0.04	-0.19	0.00
9	0.18	-0.28	0.01
10	0.10	-2.22**	0.19
11	0.06	-1.94	0.09
12	0.07	-0.50	0.01
13	0.04	-0.62	0.01
14	0.02	-0.03	0.00
15	0.02	-0.72	0.01
16	0.07	0.14	0.00
17	0.02	-0.62	0.00
18	0.04	-1.40	0.04
19	0.03	1.21	0.02
20	0.09	-0.45	0.01
21	0.01	0.16	0.00
22	0.11	0.06	0.00
23	0.43	-0.11	0.00
24	0.04	0.19	0.00
25	0.05	0.44	0.00

*Leverages >3(6 + 1)/45 = .47 were flagged

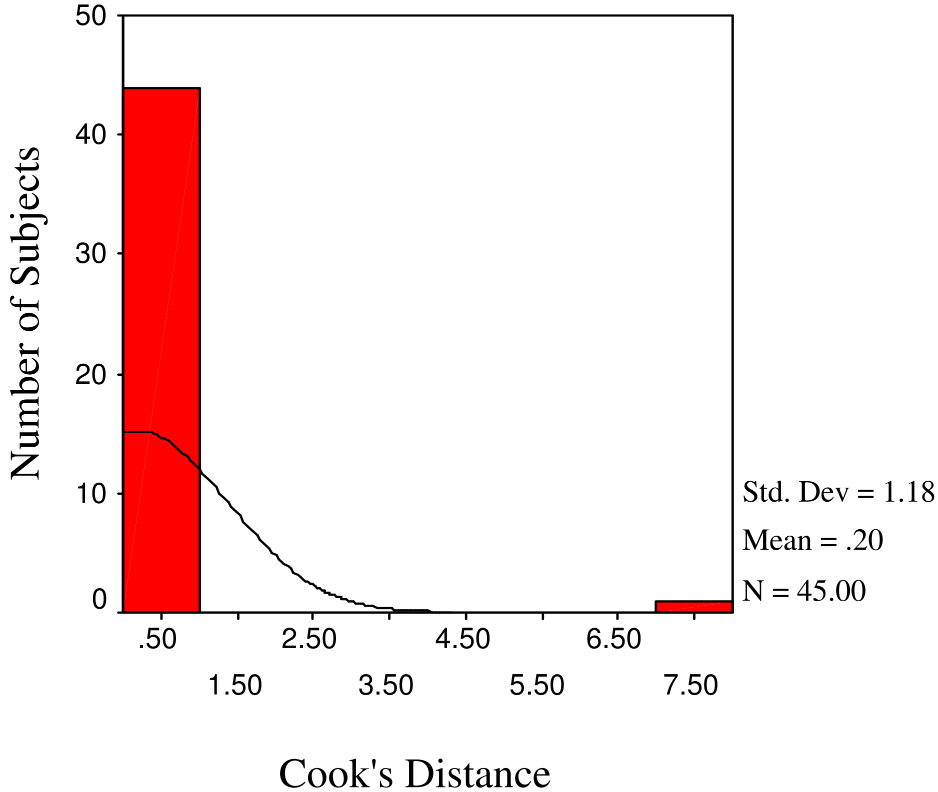
**Z-residuals >|2| were flagged

Subject	Leverage*	Z- residual**	Cook's D
26	0.01	0.27	0.00
27	0.03	-0.48	0.00
28	0.21	3.72**	1.34
29	0.03	0.81	0.01
30	0.01	0.66	0.00
31	0.07	1.19	0.04
32	0.00	0.04	0.00
33	0.12	-0.34	0.01
34	0.06	-0.42	0.00
35	0.02	0.68	0.00
36	0.01	-0.61	0.00
37	0.04	0.52	0.00
38	0.01	0.12	0.00
39	0.01	-0.46	0.00
40	0.03	1.78	0.04
41	0.06	-0.47	0.01
42	0.01	0.65	0.00
43	0.01	-0.32	0.00
44	0.46	0.14	0.01
45	0.03	-0.20	0.00

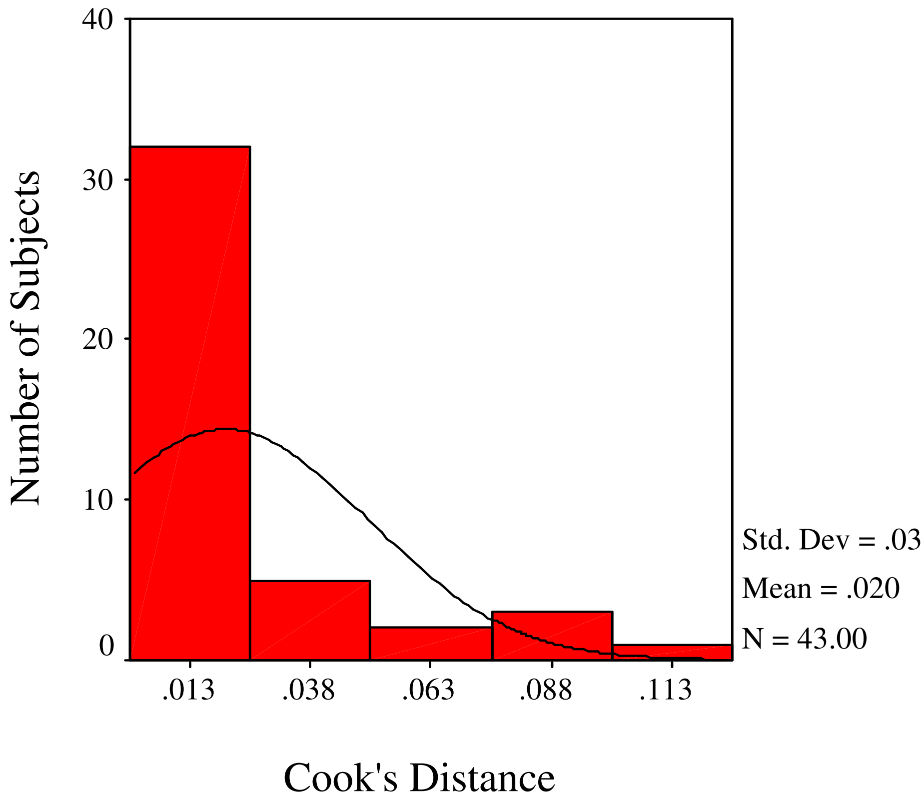
*Leverages >3(6 + 1)/45 = .47 were flagged

**Z-residuals >|2| were flagged

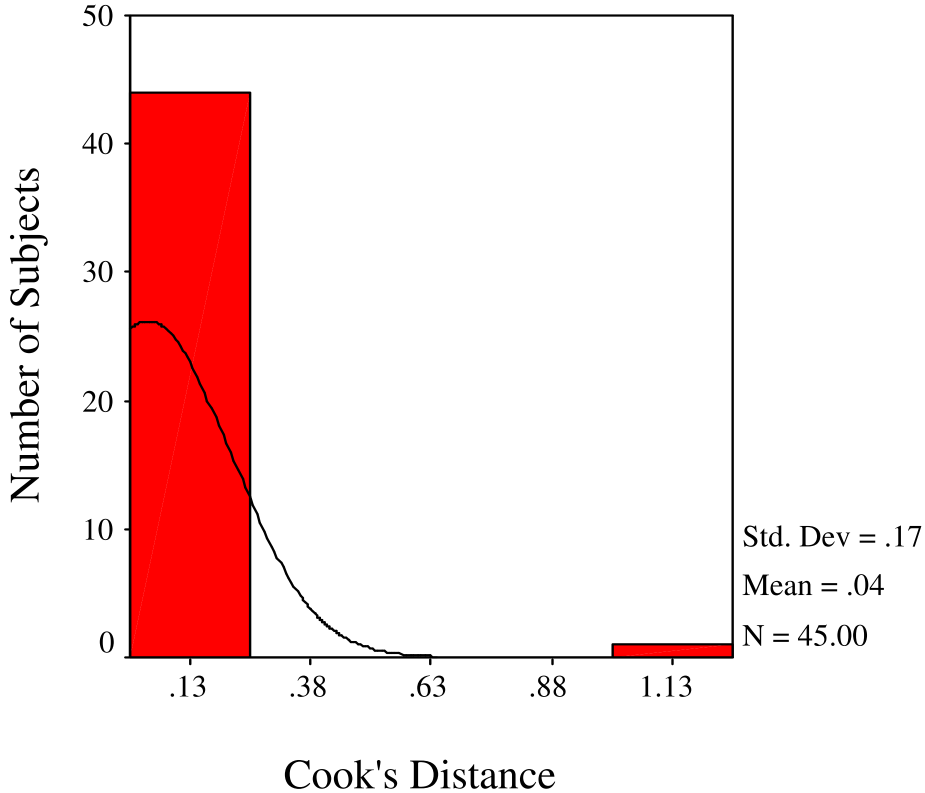
Frequency Histogram of Cook's D Values for Prediction of iAKSF – All 45 Participants



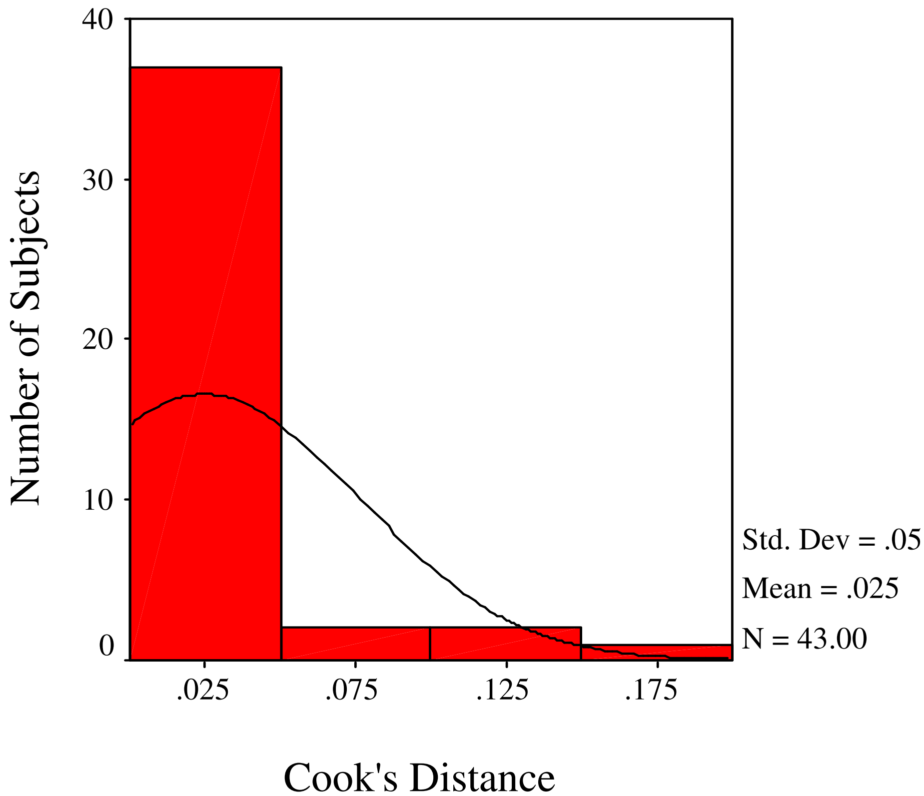
Frequency Histogram of Cook’s D for Prediction of iAKSF – Without Participants 2 and 28



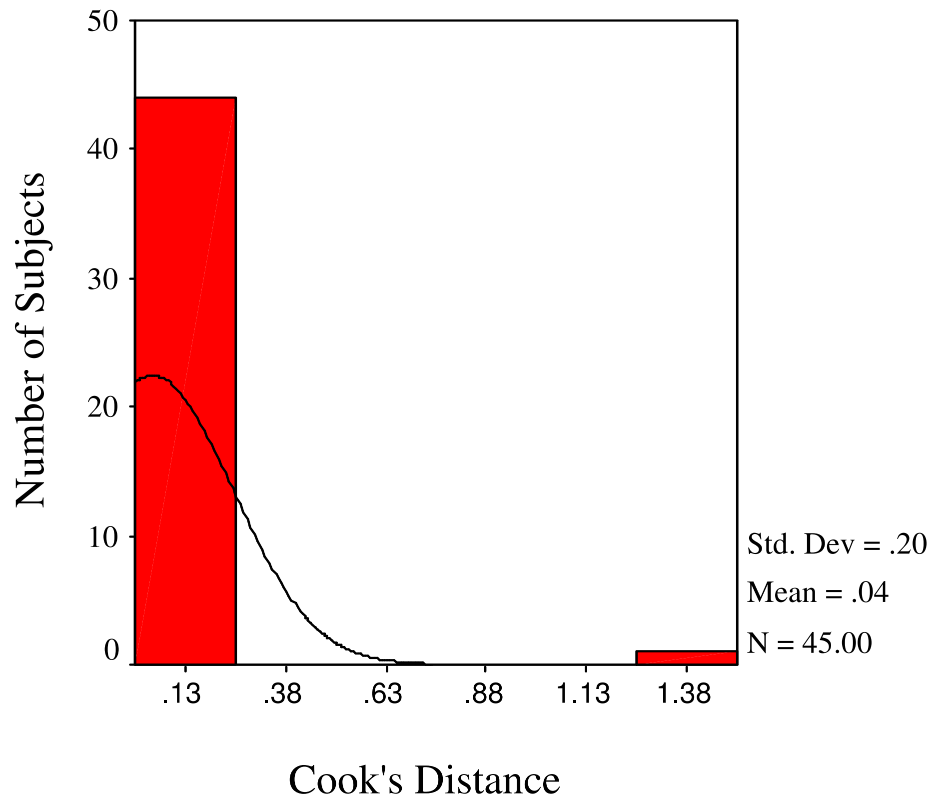
Frequency Histogram of Cook's D for Prediction of rAKSF – All 45 Participants



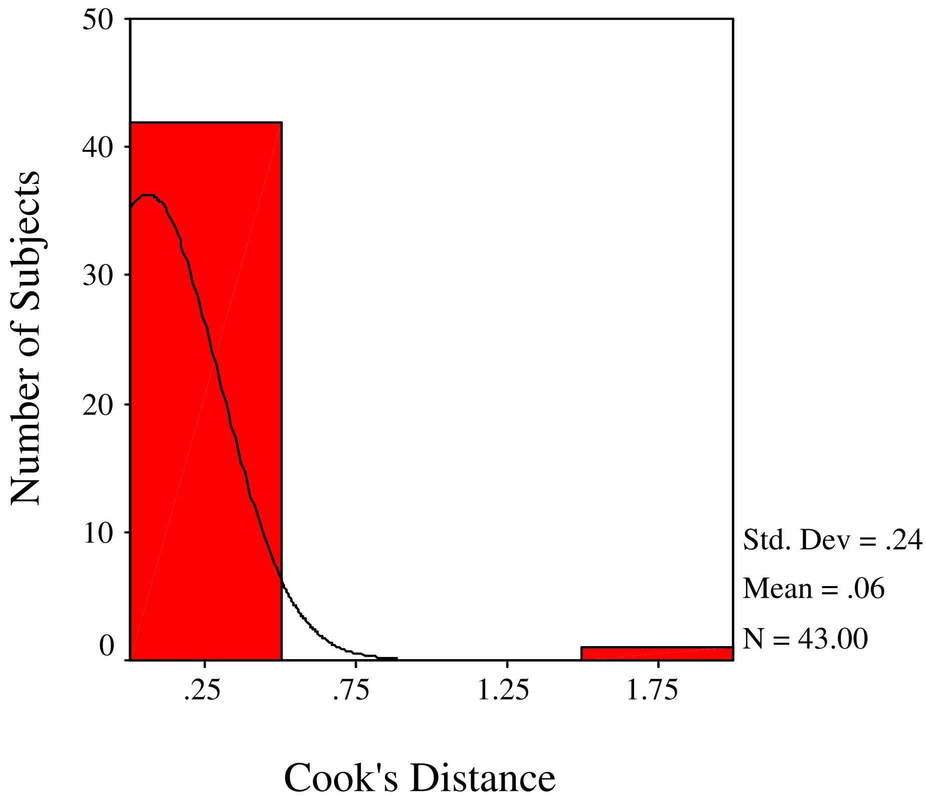
Frequency Histogram of Cook’s D for Prediction of rAKSF – Without Participants 2 and 28



Frequency Histogram of Cook's D for Prediction of pAKSF – All 45 Participants

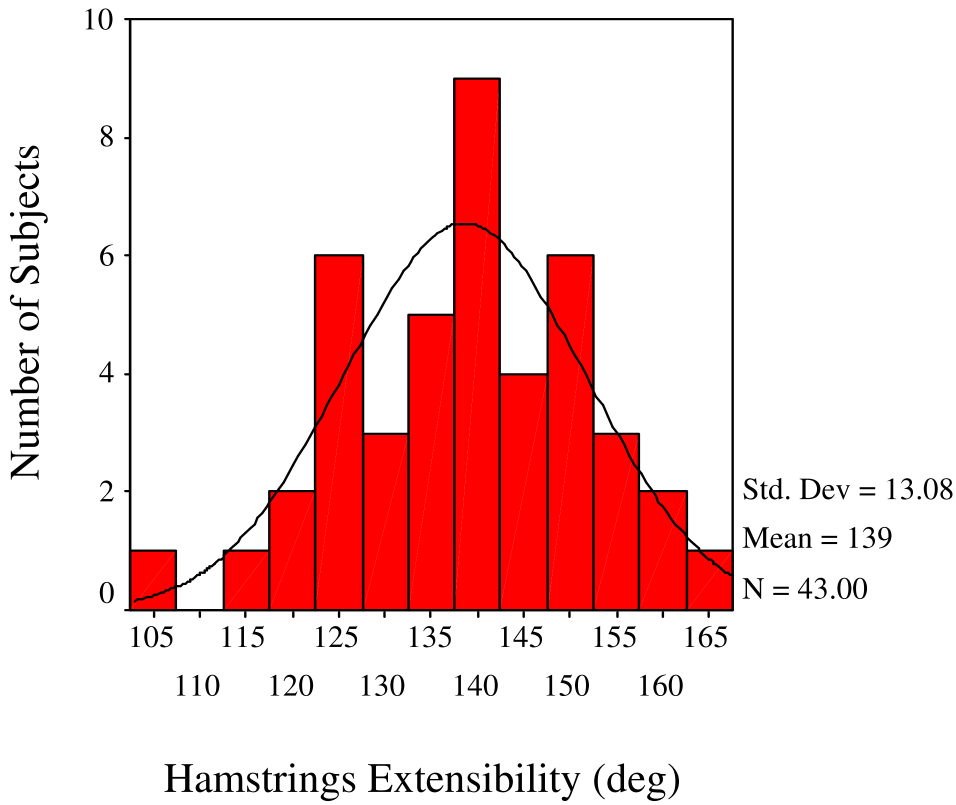


Frequency Histogram of Cook's D for Prediction of pAKSF – Without Participants 2 and 28

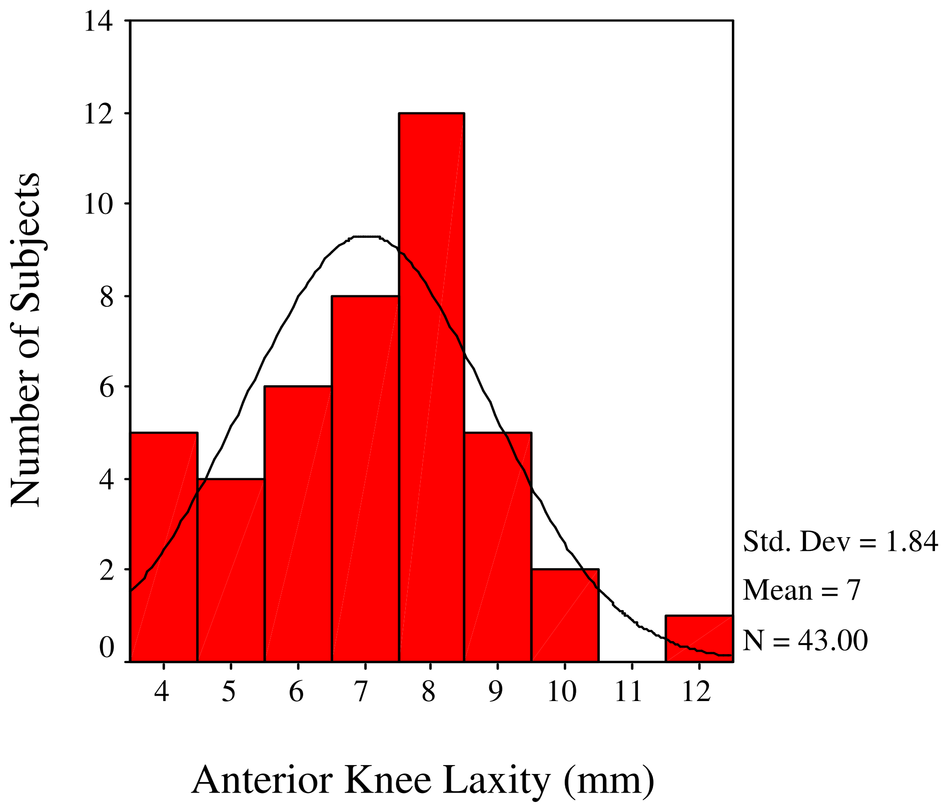


APPENDIX F_1. Predictor Variable Frequency Histograms

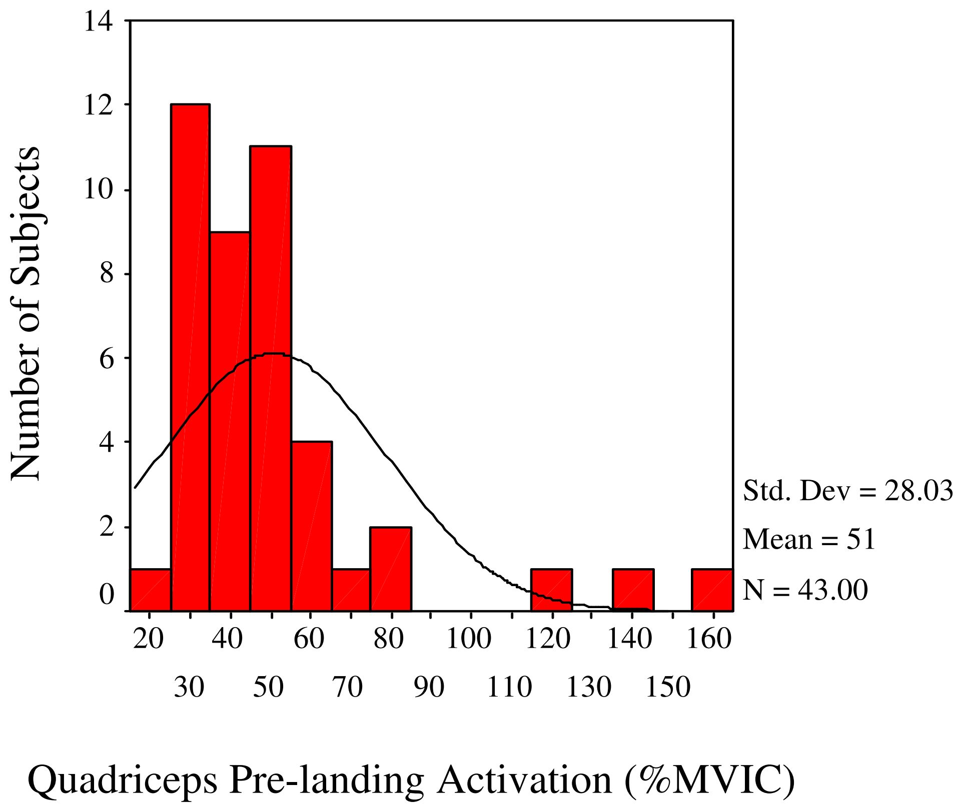
Hamstrings Extensibility Histogram



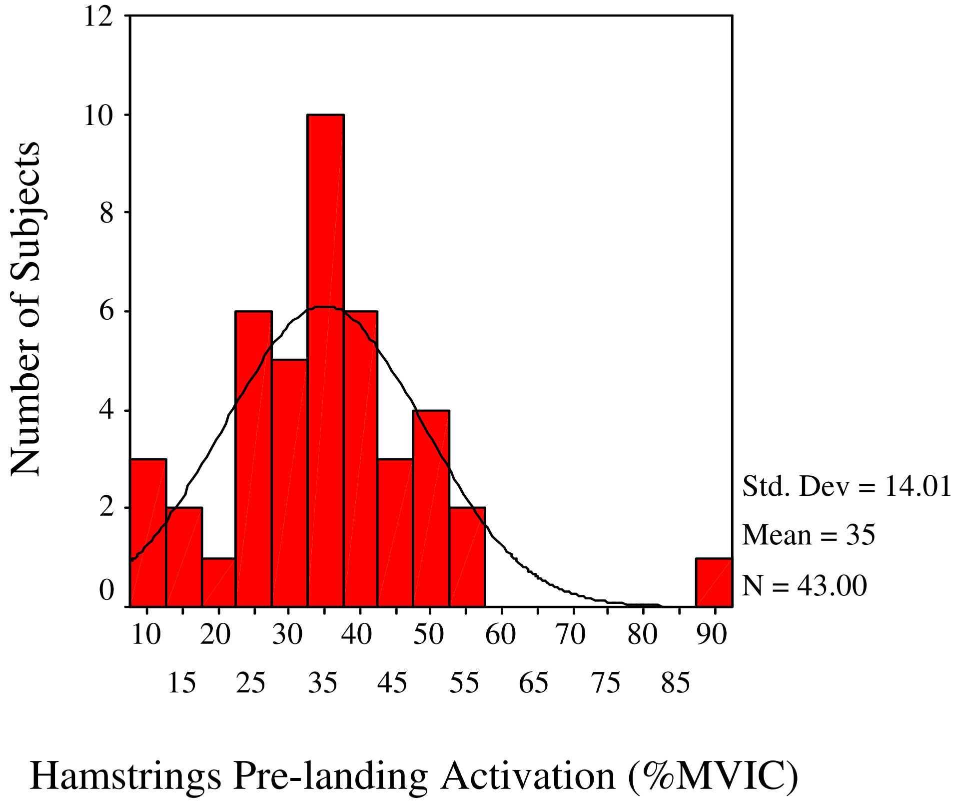
Anterior Knee Laxity Histogram



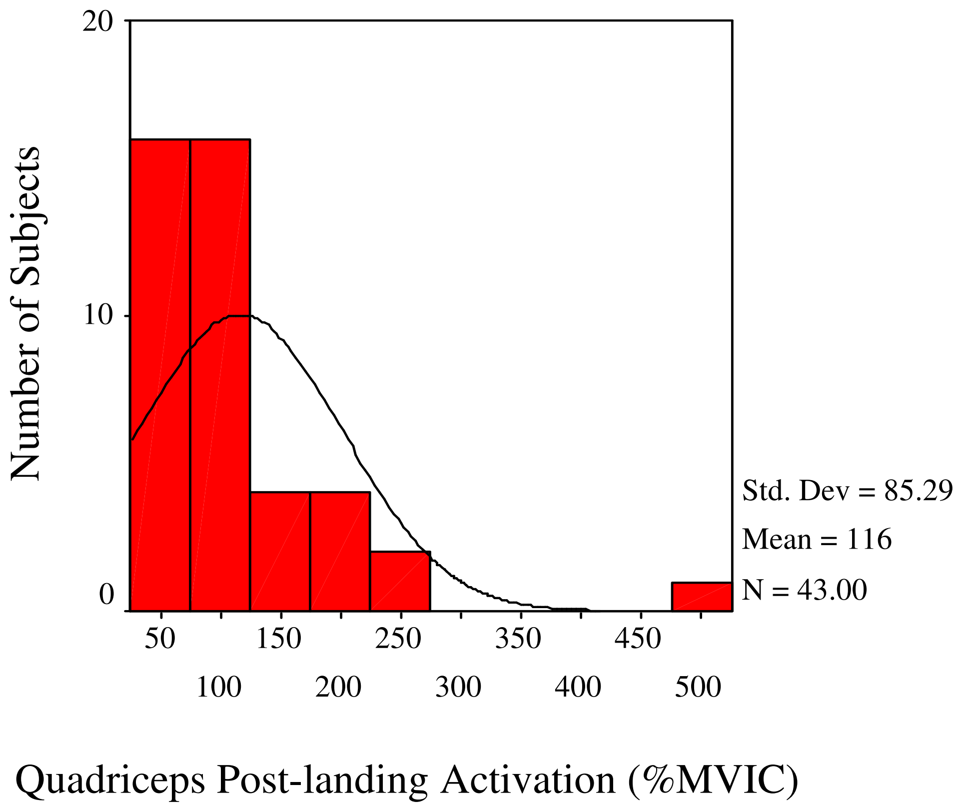
Quadriceps Pre-landing Activation Histogram



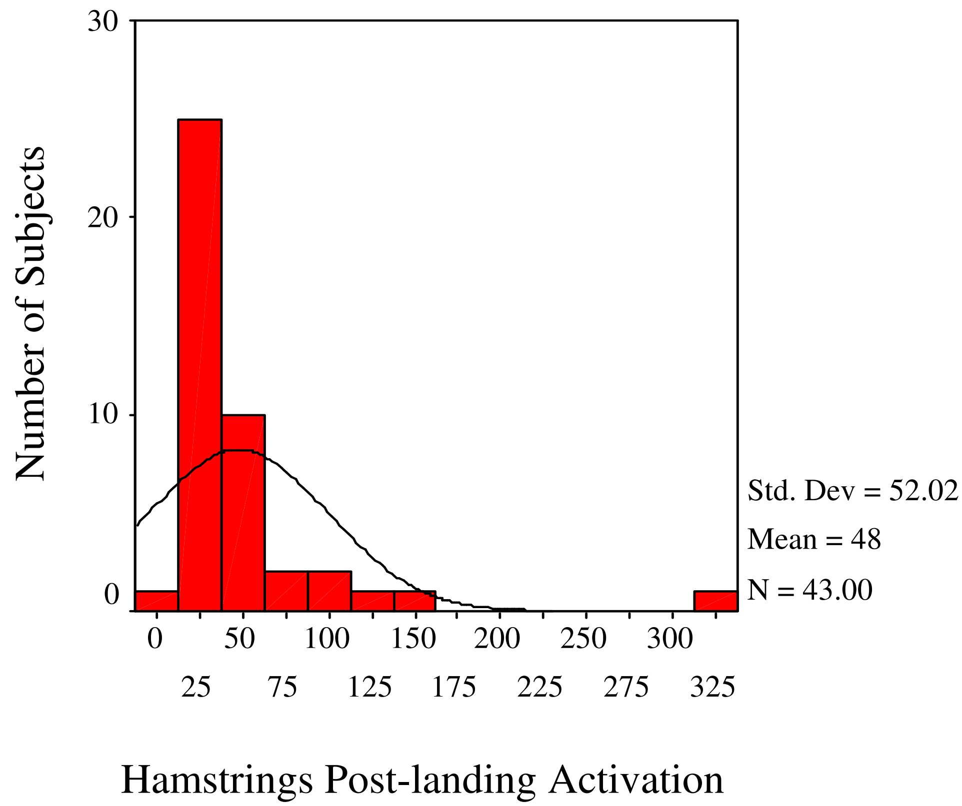
Hamstrings Pre-landing Activation Histogram



Quadriceps Post-landing Activation Histogram

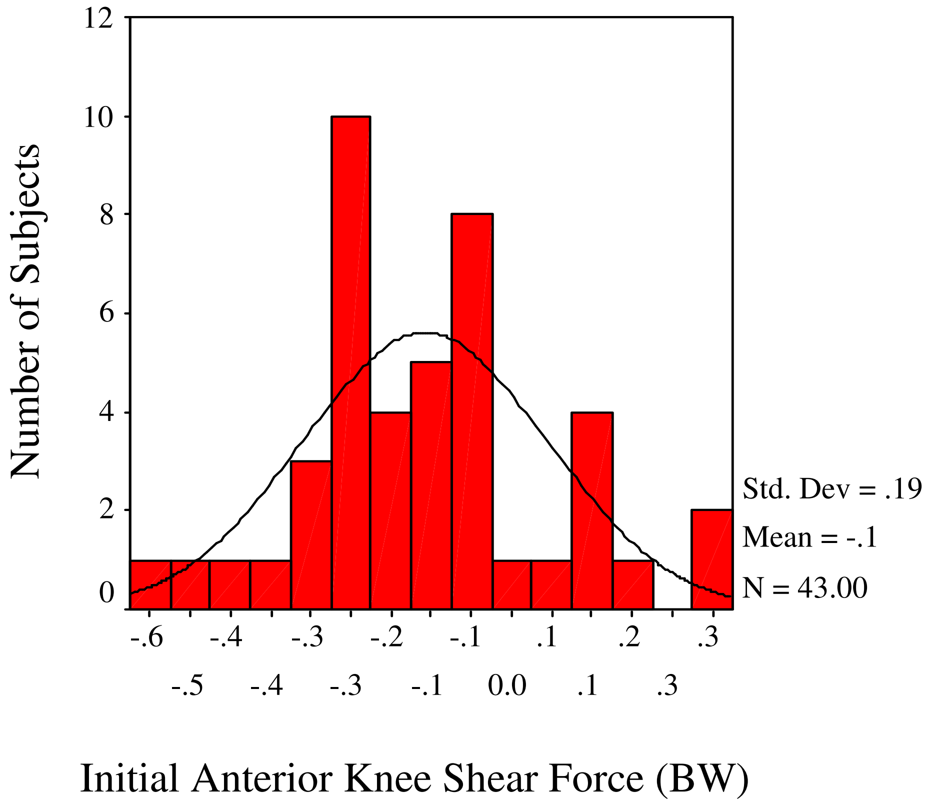


Hamstrings Post-landing Activation Histogram

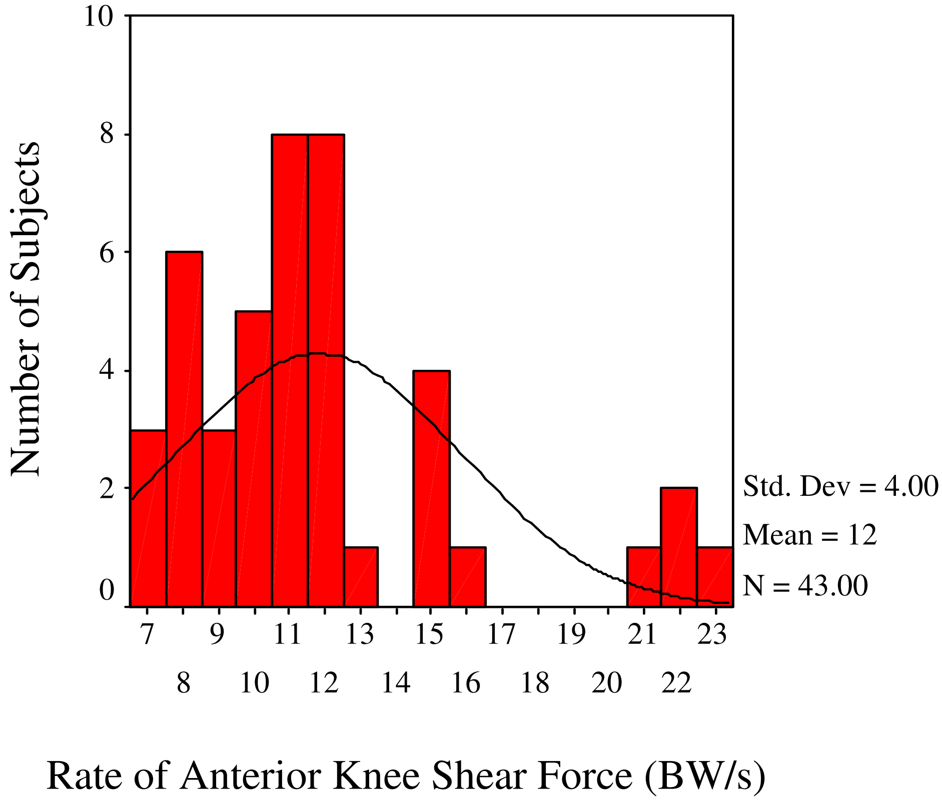


APPENDIX F_2. Dependent Variable Frequency Histograms

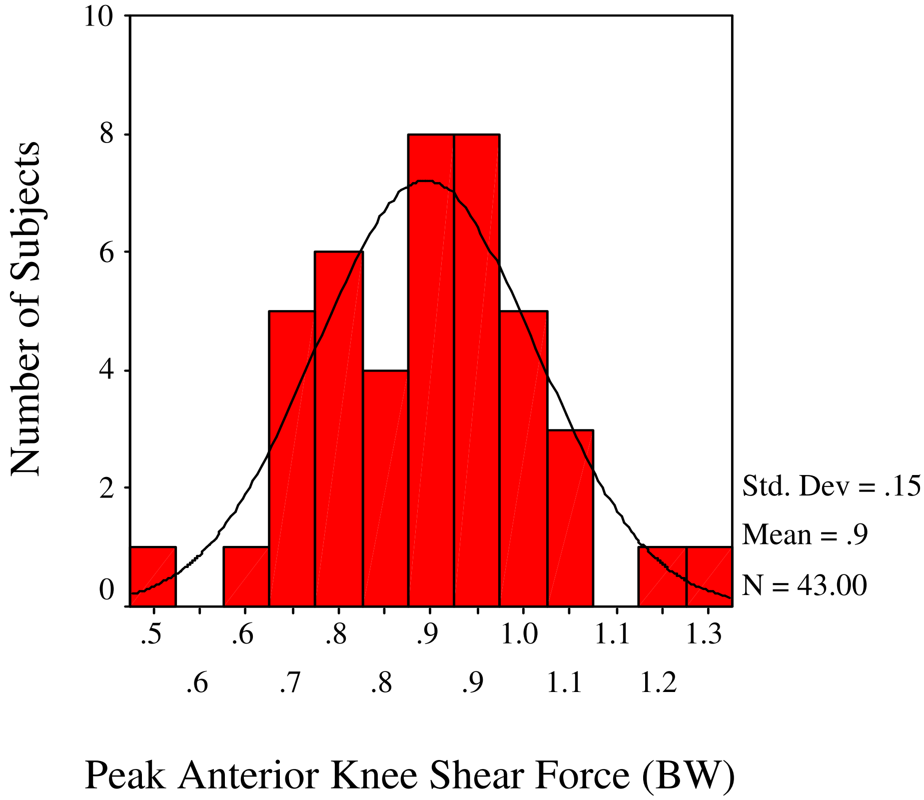
Initial Anterior Knee Shear Force Histogram



Rate of Anterior Knee Shear Force Histogram



Peak Anterior Knee Shear Force Histogram



Appendix G. Correlations, Model Summary, ANOVA, and Coefficient Tables for the Prediction of iAKSF

Correlations

		iAKSF	KT	HS	Q_{pre}	H_{pre}
Pearson Correlation	iAKSF	1	0.19	0.04	-0.31	-0.39
	KT	0.19	1	-0.10	0.14	0.06
	HS	0.04	-0.10	1	-0.18	-0.09
	Q_{pre}	-0.31	0.14	-0.18	1	0.44
	H_{pre}	-0.39	0.06	-0.09	0.44	1
Sig. (1-tailed)	iAKSF	.	0.11	0.41	0.02	0.01
	KT	0.11	.	0.27	0.18	0.34
	HS	0.41	0.27	.	0.13	0.29
	Q_{pre}	0.02	0.18	0.13	.	0.00
	H_{pre}	0.01	0.34	0.29	0.00	.
N	iAKSF	43	43	43	43	43
	KT	43	43	43	43	43
	HS	43	43	43	43	43
	Q_{pre}	43	43	43	43	43
	H_{pre}	43	43	43	43	43

Model Summary

Model	R	R^2	Adjusted R^2	Std. Error	Change Statistics				
					R^2 Change	F Change	df1	df2	Sig. F Change
1	0.389	0.151	0.131	0.178	0.151	7.318	1	41	0.01
2	0.444	0.197	0.157	0.175	0.046	2.281	1	40	0.139
3	0.481	0.232	0.172	0.174	0.034	1.743	1	39	0.194

a Predictors: (Constant), H_{pre}

b Predictors: (Constant), H_{pre}, KT

c Predictors: (Constant), H_{pre}, KT, Q_{pre}

d Dependent Variable: iAKSF

ANOVA

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	0.232	1	0.232	7.318	0.01
	Residual	1.297	41	0.032		
	Total	1.529	42			
2	Regression	0.302	2	0.151	4.914	0.012
	Residual	1.227	40	0.031		
	Total	1.529	42			
3	Regression	0.354	3	0.118	3.918	0.015
	Residual	1.175	39	0.03		
	Total	1.529	42			

a Predictors: (Constant), H_{pre}

b Predictors: (Constant), H_{pre}, KT

c Predictors: (Constant), H_{pre}, KT, Q_{pre}

d Dependent Variable: iAKSF

Coefficients

Model		Unstand Coefficients		Stand Coefficients	t	Sig.	95% CI for B		Correlations			Collinearity
		B	Std. Error	Beta			Low Bnd	Up Bnd	0-order	Partial	Part	Tolerance
1	(Constant)	0.050	0.074		0.682	0.499	-0.098	0.199				
	H_{pre}	-0.005	0.002	-0.389	-2.705	0.010	-0.009	-0.001	-0.389	-0.389	-0.389	1.000
2	(Constant)	-0.100	0.123		-0.812	0.422	-0.348	0.149				
	H_{pre}	-0.005	0.002	-0.403	-2.837	0.007	-0.009	-0.002	-0.389	-0.409	-0.402	0.996
	KT	0.022	0.015	0.214	1.510	0.139	-0.008	0.052	0.189	0.232	0.214	0.996
3	(Constant)	-0.088	0.122		-0.724	0.474	-0.335	0.159				
	H_{pre}	-0.004	0.002	-0.312	-1.993	0.053	-0.009	0.000	-0.389	-0.304	-0.280	0.804
	KT	0.025	0.015	0.238	1.678	0.101	-0.005	0.054	0.189	0.260	0.236	0.980
	Q_{pre}	-0.001	0.001	-0.208	-1.320	0.194	-0.004	0.001	-0.313	-0.207	-0.185	0.791

a Dependent Variable: iAKSF

APPENDIX H. Correlations, Model Summary, ANOVA, and Coefficient Tables for the Prediction of rAKSF

Correlations

		rAKSF	KT	HS	Qpre	Hpre	Qpost	Hpost
Pearson Correlation	rAKSF	1	-0.19	-0.09	0.13	0.41	0.19	0.38
	KT	-0.19	1	-0.10	0.14	0.06	0.35	0.14
	HS	-0.09	-0.10	1	-0.18	-0.09	-0.07	-0.10
	Q_{pre}	0.13	0.14	-0.18	1	0.44	0.70	0.56
	H_{pre}	0.41	0.06	-0.09	0.44	1	0.55	0.75
	Q_{post}	0.19	0.35	-0.07	0.70	0.55	1	0.70
	H_{post}	0.38	0.14	-0.10	0.56	0.75	0.70	1
Sig. (1-tailed)	rAKSF	.	0.12	0.29	0.21	0.00	0.11	0.01
	KT	0.12	.	0.27	0.18	0.34	0.01	0.19
	HS	0.29	0.27	.	0.13	0.29	0.32	0.27
	Q_{pre}	0.21	0.18	0.13	.	0.00	0.00	0.00
	H_{pre}	0.00	0.34	0.29	0.00	.	0.00	0.00
	Q_{post}	0.11	0.01	0.32	0.00	0.00	.	0.00
	H_{post}	0.01	0.19	0.27	0.00	0.00	0.00	.
N	rAKSF	43	43	43	43	43	43	43
	KT	43	43	43	43	43	43	43
	HS	43	43	43	43	43	43	43
	Q_{pre}	43	43	43	43	43	43	43
	H_{pre}	43	43	43	43	43	43	43
	Q_{post}	43	43	43	43	43	43	43
	H_{post}	43	43	43	43	43	43	43

Model Summary

Model	R	R^2	Adjusted R^2	Std. Error	Change Statistics				
					R^2 Change	F Change	df1	df2	Sig. F Change
1	0.412	0.170	0.149	3.686	0.170	8.380	1	41	0.006
2	0.464	0.215	0.176	3.629	0.045	2.313	1	40	0.136
3	0.484	0.234	0.175	3.630	0.019	0.977	1	39	0.329

a Predictors: (Constant), H_{pre}

b Predictors: (Constant), H_{pre}, KT

c Predictors: (Constant), H_{pre}, KT, H_{post}

d Dependent Variable: rAKSF

ANOVA

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	113.877	1	113.877	8.380	0.006
	Residual	557.148	41	13.589		
	Total	671.025	42			
2	Regression	144.331	2	72.165	5.481	0.008
	Residual	526.694	40	13.167		
	Total	671.025	42			
3	Regression	157.206	3	52.402	3.977	0.014
	Residual	513.818	39	13.175		
	Total	671.025	42			

a Predictors: (Constant), H_{pre}

b Predictors: (Constant), H_{pre}, KT

c Predictors: (Constant), H_{pre}, KT, H_{post}

d Dependent Variable: rAKSF

Coefficients

Model		Unstand Coefficients		Stand Coefficients	t	Sig.	95% CI for B		Correlations			Collinearity
		B	Std. Error	Beta			Low Bnd	Up Bnd	0-order	Partial	Part	Tolerance
1	(Constant)	7.713	1.524		5.061	0.000	4.635	10.791				
	H_{pre}	0.118	0.041	0.412	2.895	0.006	0.036	0.199	0.412	0.412	0.412	1.000
2	(Constant)	10.841	2.546		4.259	0.000	5.696	15.985				
	H_{pre}	0.121	0.040	0.425	3.031	0.004	0.040	0.202	0.412	0.432	0.425	0.996
	KT	-0.464	0.305	-0.213	-1.521	0.136	-1.081	0.153	-0.187	-0.234	-0.213	0.996
3	(Constant)	11.934	2.776		4.299	0.000	6.319	17.548				
	H_{pre}	0.076	0.061	0.267	1.255	0.217	-0.047	0.199	0.412	0.197	0.176	0.433
	KT	-0.506	0.308	-0.233	-1.641	0.109	-1.129	0.118	-0.187	-0.254	-0.230	0.977
	H_{post}	0.016	0.016	0.212	0.989	0.329	-0.017	0.050	0.381	0.156	0.139	0.427

a Dependent Variable: rAKSF

APPENDIX I. Correlations, Model Summary, ANOVA, and Coefficient Tables for the Prediction of pAKSF

Correlations

		pAKSF	KT	HS	Q_{pre}	H_{pre}	Q_{post}	H_{post}
Pearson Correlation	pAKSF	1	0.01	-0.06	0.02	0.03	0.14	0.32
	KT	0.01	1	-0.10	0.14	0.06	0.35	0.14
	HS	-0.06	-0.10	1	-0.18	-0.09	-0.07	-0.10
	Q_{pre}	0.02	0.14	-0.18	1	0.44	0.70	0.56
	H_{pre}	0.03	0.06	-0.09	0.44	1	0.55	0.75
	Q_{post}	0.14	0.35	-0.07	0.70	0.55	1	0.70
	H_{post}	0.32	0.14	-0.10	0.56	0.75	0.70	1
Sig. (1-tailed)	pAKSF	.	0.47	0.34	0.46	0.44	0.19	0.02
	KT	0.47	.	0.27	0.18	0.34	0.01	0.19
	HS	0.34	0.27	.	0.13	0.29	0.32	0.27
	Q_{pre}	0.46	0.18	0.13	.	0.00	0.00	0.00
	H_{pre}	0.44	0.34	0.29	0.00	.	0.00	0.00
	Q_{post}	0.19	0.01	0.32	0.00	0.00	.	0.00
	H_{post}	0.02	0.19	0.27	0.00	0.00	0.00	.
N	pAKSF	43	43	43	43	43	43	43
	KT	43	43	43	43	43	43	43
	HS	43	43	43	43	43	43	43
	Q_{pre}	43	43	43	43	43	43	43
	H_{pre}	43	43	43	43	43	43	43
	Q_{post}	43	43	43	43	43	43	43
	H_{post}	43	43	43	43	43	43	43

Model Summary

Model	R	R^2	Adjusted R^2	Std. Error	Change Statistics				
					R^2 Change	F Change	df1	df2	Sig. F Change
1	0.317	0.101	0.079	0.143	0.101	4.583	1	41	0.038
2	0.452	0.205	0.165	0.136	0.104	5.231	1	40	0.028
3	0.488	0.238	0.180	0.134	0.034	1.734	1	39	0.196

a Predictors: (Constant), H_{post}

b Predictors: (Constant), H_{post}, H_{pre}

c Predictors: (Constant), H_{post}, H_{pre}, Q_{pre}

d Dependent Variable: pAKSF

ANOVA

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	0.093	1	0.093	4.583	0.038
	Residual	0.833	41	0.02		
	Total	0.926	42			
2	Regression	0.189	2	0.095	5.143	0.01
	Residual	0.736	40	0.018		
	Total	0.926	42			
3	Regression	0.221	3	0.074	4.07	0.013
	Residual	0.705	39	0.018		
	Total	0.926	42			

a Predictors: (Constant), H_{post}

b Predictors: (Constant), H_{post}, H_{pre}

c Predictors: (Constant), H_{post}, H_{pre}, Q_{pre}

d Dependent Variable: pAKSF

Coefficients

Model		Unstand Coefficients		Stand Coefficients	t	Sig.	95% CI for B		Correlations			Collinearity
		B	Std. Error	Beta			Low Bnd	Up Bnd	0-order	Partial	Part	Tolerance
1	(Constant)	0.825	0.030		27.881	0.000	0.765	0.885				
	H_{post}	0.001	0.000	0.317	2.141	0.038	0.000	0.002	0.317	0.317	0.317	1.000
2	(Constant)	0.956	0.064		14.978	0.000	0.827	1.085				
	H_{post}	0.002	0.001	0.685	3.202	0.003	0.001	0.003	0.317	0.452	0.452	0.435
	H_{pre}	-0.005	0.002	-0.489	-2.287	0.028	-0.010	-0.001	0.026	-0.340	-0.323	0.435
3	(Constant)	0.996	0.070		14.204	0.000	0.854	1.138				
	H_{post}	0.002	0.001	0.802	3.489	0.001	0.001	0.004	0.317	0.488	0.488	0.370
	H_{pre}	-0.005	0.002	-0.479	-2.257	0.030	-0.010	-0.001	0.026	-0.340	-0.315	0.434
	Q_{pre}	-0.001	0.001	-0.222	-1.317	0.196	-0.003	0.001	0.016	-0.206	-0.184	0.684

a Dependent Variable: pAKSF

APPENDIX J. Variable Correlation Matrix

Bivariate Pearson Correlations

	KT	HS	iAKSF	pAKSF	tpAKSF	rAKSF	Q_{pre}	Q_{post}	H_{pre}	H_{post}
KT	1									
HS	-0.10	1								
	0.53									
iAKSF	0.19	0.04	1							
	0.23	0.82								
pAKSF	0.01	-0.06	-0.02	1						
	0.93	0.68	0.91							
tpAKSF	0.04	0.08	-0.21	0.03	1					
	0.78	0.62	0.18	0.87						
rAKSF	-0.19	-0.09	-0.46	0.44	-0.62	1				
	0.23	0.58	0.00*	0.00*	0.00*					
Q_{pre}	0.14	-0.18	-0.31	0.02	0.06	0.13	1			
	0.37	0.25	0.04*	0.92	0.71	0.43				
Q_{post}	0.35	-0.07	-0.26	0.14	-0.05	0.19	0.70	1		
	0.02*	0.64	0.10	0.39	0.75	0.22	0.00*			
H_{pre}	0.06	-0.09	-0.39	0.03	-0.23	0.41	0.44	0.55	1	
	0.69	0.58	0.01*	0.87	0.14	0.01*	0.00*	0.00*		
H_{post}	0.14	-0.10	-0.31	0.32	-0.08	0.38	0.56	0.70	0.75	1
	0.38	0.54	0.04*	0.04*	0.61	0.01*	0.00*	0.00*	0.00*	

*Value is Significant at alpha = .05